

**A biomechanical investigation into the link between
simulated job static strength and psychophysical strength:
Do they share a “weakest link” relationship?**

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

Maximum voluntary forces and psychophysically acceptable forces are often used to set force guidelines for exertions as a means to protect against overexertion injuries in the workplace. The focus of this dissertation was the exploration of the roles of whole body balance, shoe-floor friction and joint strength in limiting the capacity of a person to produce maximum voluntary hand forces and psychophysically acceptable hand forces. The underlying goal was to advance knowledge regarding how physical exertion capacity is biomechanically governed, then to use this information to develop models to predict capability based on these governing principles. The hypothesis underscoring this work was that maximum voluntary hand force capability is governed by whole body balance, shoe-floor friction and joint strength; and consequently, psychophysically acceptable forces would be chosen proportionally to this maximum voluntary force capability, where the magnitude of the proportionality was dependent on the limiting factor, or 'weakest link'.

To investigate this hypothesis, both experimental and mathematical modeling paradigms were used. Initially, an experimental study was used to investigate how biomechanical factors governed maximum hand force capability across a range of exertions. It revealed that each governing factor differentially limited maximum force capability. Moreover, this study identified how foot placement, handle height, distance from the handle, friction, and body posture all influence the underlying biomechanical weakest link, and ultimately force producing capability.

Data gathered in the experimental study was next used to evaluate a mathematical model that was developed to predict maximum force capability, given information on posture and direction of force application. In addition, the model also predicted population variability

in maximum capacity based on the inclusion of a novel approach to probabilistically represent population variability. The evaluation demonstrated that the model underestimated maximum hand force capability compared to measured hand forces by approximately 18, 26, and 41% during medial, pulling and downward exertions respectively. However, it appeared that the ‘weakest link’ principle for predicting maximum force capacity was plausible, as evidenced by significant rank ordered correlations between the measured and predicted hand forces.

Further research investigated if psychophysically acceptable forces were selected as a proportion of task specific maximum voluntary force capability, where the proportionality was related to the biomechanical weakest link. Using an experimental design, psychophysically acceptable forces and corresponding maximum forces were measured. Participants chose psychophysically acceptable forces that were $4/5^{\text{th}}$ of their task specific maximum voluntary force capability when capability was limited by balance. Additionally, they chose psychophysically acceptable forces that were $2/3^{\text{rd}}$ of their maximum voluntary force capability when capability was limited by joint strength. The identification and confirmation of a weakest link proportionality principle represents an important contribution to the field of occupational biomechanics.

The weakest link proportionality principle was integrated into the model to allow prediction of: maximum voluntary hand force capability, the limiting factor, and psychophysically acceptable hand force capability. The updated model underestimated empirically measured psychophysically acceptable forces by 24% and 43% during downward and pulling exertions respectively. However, the original model underestimated the maximum hand force capacity by 23% and 34% during the same exertions, without the proportional relationships. This underestimation may be a result of the underlying assumption that joint

strength is independent, resulting in an underestimation of maximum joint strength capacity and a corresponding underestimation of maximum hand force capacity. The underestimation may also be due to differences in strength capacities between the participants tested during this thesis compared to those tested in past research used to determine the maximum strength indices reported in the literature.

This body of work supported the hypothesis that psychophysically acceptable forces are selected as a proportion of the maximum voluntary hand force, where the proportionality depends on the underlying biomechanical weakest link. The model is a promising first step towards predicting maximum and psychophysically acceptable occupational force threshold limits.

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Chapter 1 – General Overview

The focus of this dissertation was to explore work capacity from the perspective of the worker as a means to provide evidence based exposure guidelines for sub-maximal work. More specifically, this research aimed to quantify relationships between maximum work capacity, defined as a worker's upper limit to produce force for a given workplace task and subjectively determined sub-maximal capacity, defined as the maximum *acceptable* amount of force produced for a repetitive task. The general hypothesis investigated in this thesis is that workers choose to perform sub-maximal repetitive work at a certain percentage of their maximum work capacity, where this percentage is related to the biomechanical component limiting their maximum capacity for the given work conditions, whether it is individual joint strength, whole body balance, friction, or another alternative. This framework is built upon the assumption that a worker will not choose to work at a level that could result in injury. Although this assumption has not been validated (Snook, 1999) this approach is useful for providing guidelines for work in the absence of a more complex rule based on a presently unachieved comprehensive understanding of the multi-factorial interactions that occur and result in the development of a disorder.

A work capacity focus was used within the paradigm of strength. The strength paradigm provides a means to describe and classify both maximum and sub-maximum capabilities and provide a backdrop for understanding capacity and demand in the context of this thesis.

1.1 – The paradigm of strength in occupational biomechanics

Muscular strength has received a great deal of attention in occupational biomechanics and ergonomics as an indicator of workplace performance capability. Dating back to the 19th

century, researchers have used muscular strength as a metric of “all-around ability” by testing individual maximum capable push, pull, or lift forces (Sargent, 1897). Remarkably, over 100 years later, this approach still persists as a popular method for assessing workplace performance capacity (Mital & Kumar, 1998 a; b). The paradigm continues to evolve over time as advances are made in both strength testing equipment and our understanding of the characteristics of strength.

Muscular strength can be defined in terms of the characteristics of the exertion as well as the characteristics of the strength application. Mital and Kumar (1998a; 1998b) provide a description of each form of muscular strength as it relates to work performance and capacity (Figure 1.1).

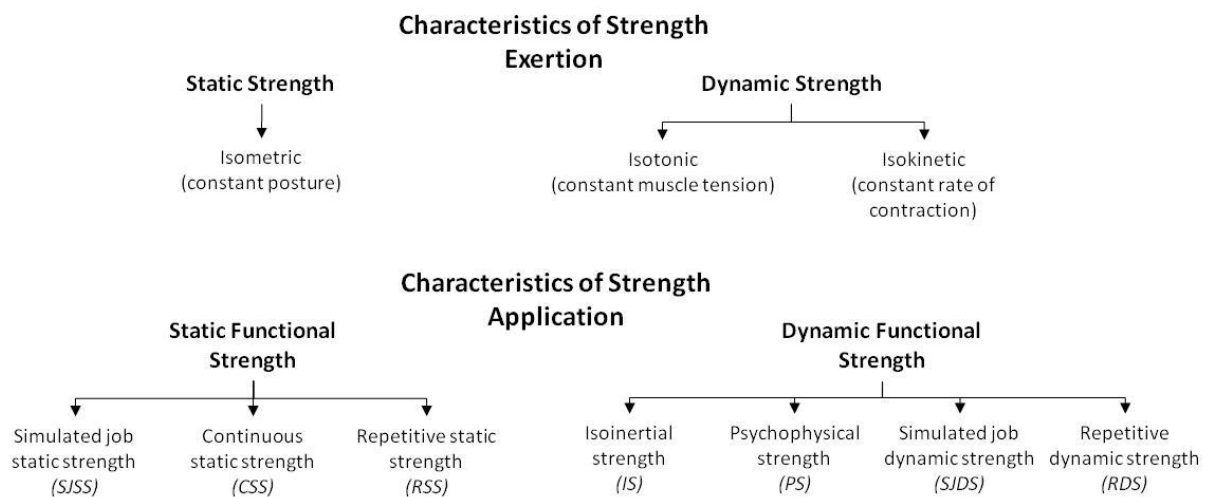


Figure 1.1 – Classification and characteristics of strength exertions and applications (as described by Mital & Kumar, 1998).

Each form of strength has been scrutinized and researched within the occupational biomechanics paradigm to determine how it relates to human capability and injury. Of the strengths indicated in Figure 1.1, three characteristics of muscular strength application have been transitioned into useful applied ergonomic assessment tools. These characteristics are: simulated job static strength, psychophysical strength, and simulated job dynamic strength.

Simulated job static strength and psychophysical strength based applied ergonomics tools have quickly been adopted as the most used of any ergonomics risk assessment tool (Dempsey et al., 2005). Simulated dynamic job strength is more often used as a screening tool to determine candidate readiness for a job (Harbin and Olson, 2005), rather than as a risk assessment based tool. Each of these measures provides useful information that can be used to monitor a worker's capacity. This dissertation endeavored to expand the usefulness of simulated job static strength and psychophysical strength in ergonomic applications.

1.2 – Simulated job static and psychophysical strength applied to ergonomics assessment

Simulated job static strength can be measured experimentally, or predicted using ergonomic software applications. This type of strength is defined as the maximum voluntary force (MVF) that can be produced within the constraints of the task being simulated. The most common prediction software used in ergonomics is the Michigan 3D Static Strength Prediction Program (3DSSPP). With this software, users are able to configure a mannequin into a specified work posture and include the magnitude and direction of the applied hand force to simulate a work task. Using these inputs, a biomechanical link segment and joint model then provide an estimate of the individual joint strengths required to complete that task, referenced to population strength. The reference to population joint strengths provides an index of the task requirements relative to maximum strength capability. This form of output is appealing for ergonomists. By modifying task requirements accordingly, an ergonomist can ensure that a minimum percentage of the workplace's population could perform the task. The software has continuously progressed since its conceptual origin (Martin & Chaffin, 1972), evolving theoretically and computationally based on emerging research evidence; a key strength of the program. This dissertation aimed to further advance the usefulness of simulated job static

strength, extending from the strong research and conceptual framework of the 3DSSPP approach.

Psychophysical strength is commonly used in ergonomic applications to determine sub-maximal work capabilities during repetitive or cyclic tasks. It refers to the psychophysically acceptable force (PAF) that can be produced repetitively within the constraints of the task being simulated, without any undue pain or fatigue. Many other applications of psychophysics have also been used in ergonomics applications such as rate of perceived exertion and so on; however, the focus of this work remains on the application of psychophysical force estimation as a means to determine sub-maximal exertion thresholds. Its application to ergonomics stems primarily from work by Dr. Stover Snook and his colleagues at Liberty Mutual Insurance. Through decades of psychophysical research they developed the Liberty Mutual Manual Materials Handling Guide, also referred to as the Snook tables (Snook, 1978; Snook and Ciriello, 1991). The fundamental benefit in using psychophysical strength within the context of ergonomics is its ability to provide threshold values for repetitive sub-maximal work. As physical labor continues to be dominated by low-force highly repetitive work there is an increasing demand to incorporate risk assessment tools that focus on these types of efforts (Marras et al., 2009).

The application of psychophysical strength in ergonomics is attractive. Using a look-up-table approach, an ergonomist can find the population sub-maximal capability for a task as defined by the primary action, the work height, the frequency, and the hand distance. The resulting table value then serves as the threshold for comparison against current task requirements. If the requirements exceed the value listed in the Liberty Mutual Manual Materials Handling Guide, the task should be modified until the exposures no longer exceed

the listed value. The simplicity and ease of use of this method is favored by ergonomists (Pascual and Naqvi, 2008). The challenge with psychophysical limits is their dependency on a myriad of work factors including posture, force application direction, friction, etc. Therefore, each psychophysically derived threshold is specific to those dependencies under which it was derived. In order to advance the usefulness and range of application for psychophysical strength, we must try to explore if underlying biomechanical based principles can be useful in helping to understand how these factors interact to dictate psychophysical capacity. Hence, a secondary thrust of this dissertation attempted to provide a more quantitative rationale to explain the influence of these factors in an effort to increase our capacity to predict psychophysical strength. An improved predictive capacity can benefit the ergonomics community by increasing the specificity of the assessments, furthering the usefulness of the psychophysical approach.

1.3 – Contrasting simulated job static strength and psychophysical strength

Simulated job static strength and psychophysical strength have opposing attributes making them each suited to differing assessment situations. Using 3DSSPP as an example, simulated job static strength is concerned with gauging actual task requirements relative to the maximum capacity – a peak approach. Psychophysical strength does not assess task requirements relative to maximum capacity. Rather, it focuses on the maximum *acceptable* capability, where acceptability is generally defined as a sub-maximal level of force that does not result in any undue pain, discomfort, or fatigue – a sub-maximal approach. While simulated job static strength is dependent primarily on posture and force direction, psychophysical strength capability is additionally dependent on repetition and rest.

From a biomechanical perspective, it is intriguing to consider the causes of the differences that exist between simulated job static and psychophysical strengths. Recent evidence proposes that these two modes of strength may share a predictable relationship (Nussbaum and Lang, 2005; Potvin, 2007). Hypothetically, psychophysical strength may be chosen, in part to maintain a specific margin of safety at the mechanical link that limits the simulated job static strength for that task. Therefore any relationship between psychophysical strength and simulated job static strength hinges on identifying the mechanical weakest link. Taking this information into consideration, the third aim of this dissertation was to examine the plausibility of a weakest link relationship between simulated job static and psychophysical strengths.

1.4 – Thesis statement

This dissertation has a central thesis: psychophysical strength is quantitatively related to simulated job static strength during one handed exertion tasks. The quantitative relationship will be related to the exposure at the joint or element most likely to limit maximum capacity during a task.

1.5 – Thesis aims and objectives

1. To advance our capacity to predict simulated job static strength by developing a biomechanically driven model to predict maximum hand forces during unilateral exertions.
2. To predict the element, or “weakest link”, most likely to limit simulated job static strength during a unilateral exertion.
3. To investigate how psychophysical strength may be determined within the strength paradigm.

4. To examine the plausibility of a weakest link explanation for the relationship between simulated job static strength and psychophysical strength.
5. To determine the feasibility of developing an ergonomic assessment tool combining the benefits of both simulated job static strength and psychophysical strength approaches (Figure 1.2).

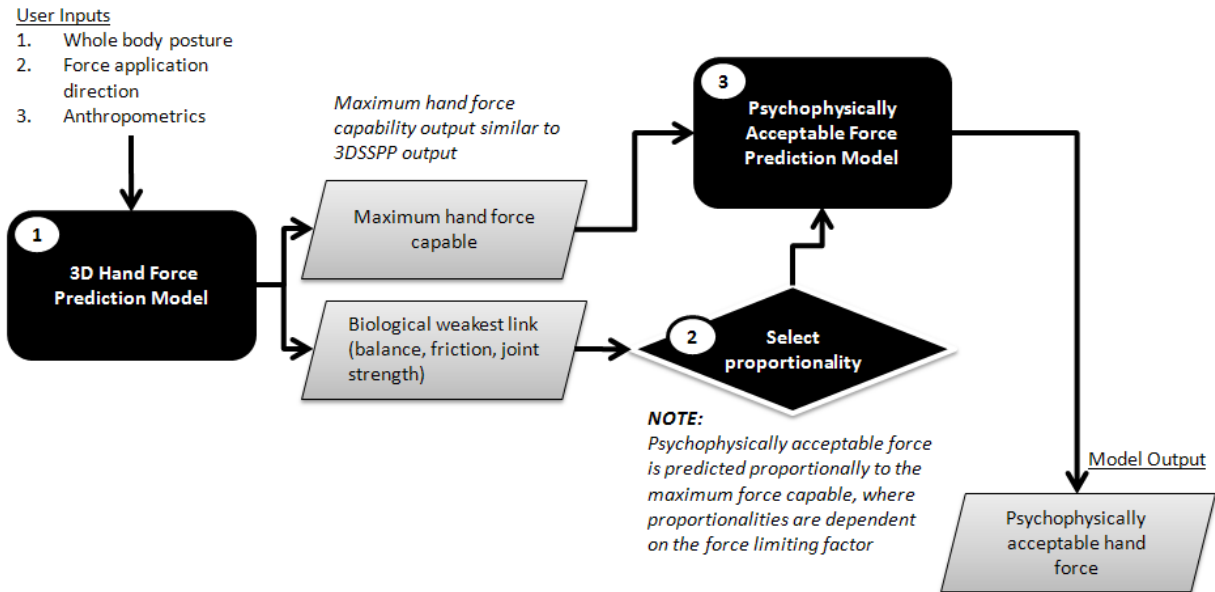


Figure 1.2 – A conceptual overview of the components and models developed throughout this dissertation.

1.6 – Thesis organization

This dissertation was written as a collection of manuscripts, placed between two introductory and one concluding chapters. All of the experimental work is dedicated to the study of one-handed simulated work tasks.

- Chapter 2 provides a review of background literature to strengthen the rationale behind the global thesis. The second chapter also serves as a discussion of the methodological considerations that guided the experimental protocols used during the experiments.

- Chapter 3 describes the experimental study examining the impact of potential limiting elements on hand force production. The experiment was conducted a) to examine how friction, balance, and joint strength influence hand force capacity, and b) to provide experimental data to be used in evaluating the 3D Hand Force Prediction Model (3DHFP). The description and evaluation of that model are provided in Chapters 4 and 5.
- Chapter 4 details the motivation and describes the model. The fourth chapter also describes how a novel stochastic approach was used to predict the weakest link limiting hand force capability.
- Chapter 5 offers an evaluation and discussion of the predictive capability of the model.
- Chapter 6 describes an experimental study that measured both simulated job static and psychophysical strength. The resulting data was used to investigate the relationships between simulated job static strength and psychophysical strength.
- Chapter 7 provides the final research piece for this dissertation by describing the feasibility of predicting psychophysical strength from simulated job static strength. Specific relationships are presented and discussed.
- Chapter 8 summarizes the major conclusions with respect to the global thesis and objectives and discusses the feasibility of using the 3DHFP in concert with a psychophysical prediction methodology to advance ergonomic assessment.

This final chapter concludes with an overview of the possible future research that could help evolve the 3DHFP into a viable, practical, ergonomics assessment tool.

Chapter 2 – Background Literature

2.1 - Overview

This review of literature is intended to provide the context and rationale behind the global thesis of this work and is presented through the following five sections.

1. Musculoskeletal disorders in the workplace – A philosophical perspective
2. Musculoskeletal disorder in the workplace – The scope of the problem
3. Common metrics to assess musculoskeletal disorder risk factors in the workplace
4. The use of psychophysical strength based metrics
5. Methodological considerations for assessing psychophysical strength

2.2 – Musculoskeletal disorders in the workplace - A philosophical perspective

Musculoskeletal disorders (MSDs) present a tremendous challenge for the workplace.

Dating back as far as the 18th century, we have known about the potential to develop disorders from work.

“The maladies that afflict clerks aforesaid arise from three causes: First, constant sitting, secondly the incessant movement of the hand and always in the same direction, thirdly the strain on the mind from the effort not to disfigure the books by errors ...” (Ramazzini, 1964: pg. 423).

“... they have to exert much effort when they use the right foot to drive round and round the larger wooden wheel ... their arms and hands are continually on the stretch as they work and hence they incur intense fatigue... So far as I can see the only sort of precaution that might help them is to avoid excess and take rest from this sort of work for several hours at a time; they should consider health more valuable than the money they make.” (Ramazzini, 1964: pg. 445-447).

Building from Ramazzini's foundational insight, researchers have paid much attention to furthering our understanding of MSD development. As a result, we have a much better understanding of how disorders may arise, and an insight into various risk factors that may initiate or exacerbate the development of an MSD (Bernard, 1997). However, much of the

current biomechanical knowledge is related to maximum tolerance levels (Marras et al. 2009), which is better suited to limiting overexertion, for example not lifting more than 23 kg (Waters et al., 1993). There is an ongoing struggle to understand injury etiology in work that involves lower levels of exertion, albeit with higher repetition, or sub-maximal work maintained for longer durations. More research is warranted to understand this complex relationship between these factors (Marras et al., 2009) and it is imperative to consider how rest can be integrated in work designs to reduce the risk of developing an MSD. Arguably, Ramazzini's sixteenth century perspective may still provide the best prevention advice, whereby workplaces “should consider health more valuable than the money they make”. In this case, workers would rest when they feel undue pain or discomfort and resume work when they feel ready.

Unfortunately, this philosophy is not likely to be adopted in today's business culture and we are forced to find alternative methods to control, reduce, or limit exposures in an effort to reduce MSD risks to acceptable levels.

2.3 –Musculoskeletal disorders in the workplace – The scope of the problem

Musculoskeletal disorders (MSDs) are a concern in the workplace. In Ontario 45.7 % of lost time injury claims were a result of overexertion, bodily reaction, repetitive motion, and static posture (WSIB, 2008). These injuries carry a heavy economic burden (Spengler et al.1986; Webster and Snook 1990; 1994a; 1994b; Silverstein et al., 1998; Punnett et al., 2000). Further, jobs with the highest injury and claims rates are often characterized as highly repetitive or requiring manual labour (Silverstein et al., 1998). In addition to the financial impact, productivity in the workplace and quality of life beyond the workplace are both likely affected (Burton et al., 2005; Ricci et al., 2006). Occupational biomechanics remains as a

fundamental approach to help continue to address MSD concerns in the workplace (Chaffin, 2009).

In order to reduce MSDs it is important to understand what factors affect their initiation or progression. Although specific injuries are categorized by specific risk factors, in a broad sense, the following six risk factors share an epidemiological link to MSDs: forceful exertion, repetitive work (frequency and duration of motion), sustained or static contraction, posture (whole body and joint specific), and vibration (Bernard, 1997). Over- or prolonged exposure to one or more risk factors can lead to MSD development (Kumar, 2001). From a conceptual perspective Armstrong and colleagues (1993) provided a parsimonious dose-response model (Figure 2.1) to demonstrate how exposure to these risk factors can impact MSD development.

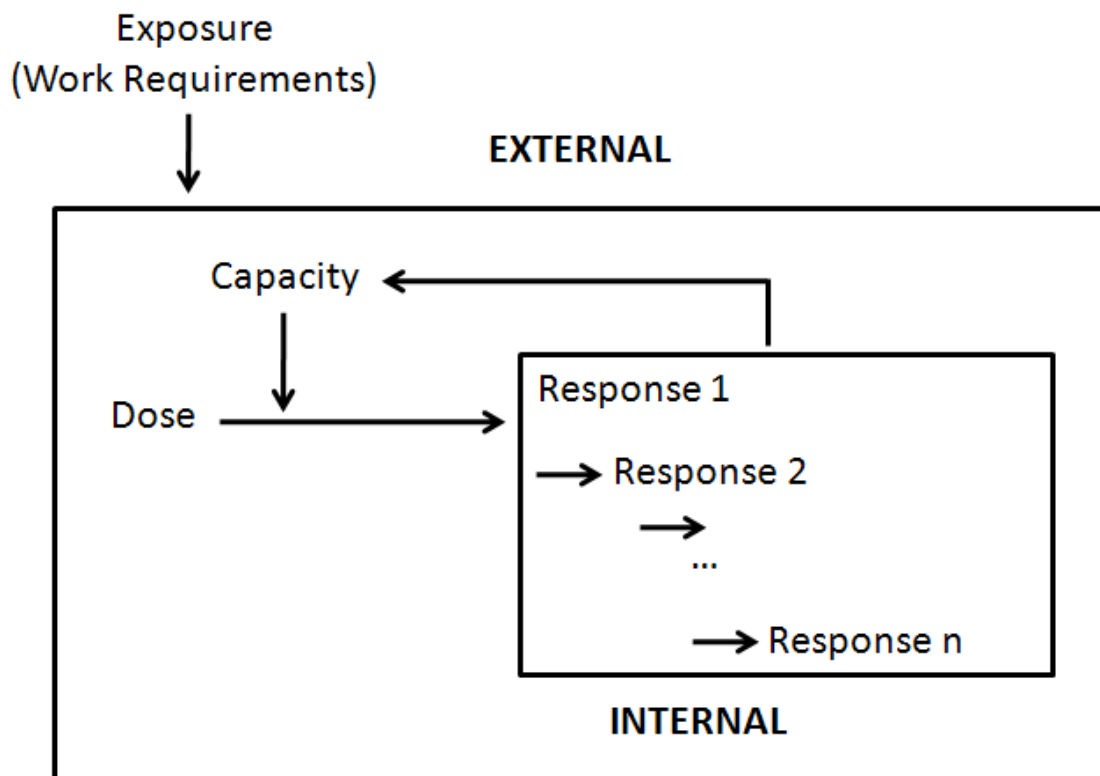


Figure 2.1 – A conceptual dose-response model to demonstrate how external exposures cause internal doses; which can result in a cascade of changes that can reduce or impair capacity, resulting in a MSD (adapted from Armstrong et al., 1993).

This model demonstrates how an exposure may affect internal tissues to varying degrees, which may ultimately modify the capacity to withstand subsequent exposures. In the context of this dissertation, this model provides a conceptual backdrop for investigating if workers subjectively select exposure levels (force) based on the corresponding internal exposure magnitude (joint strength demand) and capacity (joint strength capacity).

2.4 – Common metrics to assess musculoskeletal disorder risk factors in the workplace

Eliminating or reducing exposure to MSD risk factors in the workplace can be an effective approach to reduce or eliminate MSDs. Based on the model presented in Figure 2.1, the ideal exposure would cause an internal dose that does not initiate a negative cascade of responses, leading to a reduction in capacity or impairment. However, identifying the ideal level of exposure is not trivial, especially when risk factors interact (Bernard, 1997; Kumar, 2001), jobs are complex, and other factors workplace factors influence tasks (Bongers et al., 1993). In the absence of safe exposure limits, the focus remains on screening for, and reducing exposure doses using pencil and paper based checklists and methods. When possible, more advanced approaches including biomechanical analysis and strength assessment can also provide useful information for determining exposure limits. Additionally, physiological or other metabolic based thresholds can be used to establish guidelines for work scenarios requiring high physical demands. The remaining discussion will focus on: pencil and paper methods, biomechanical analysis, and strength assessment.

2.4.1 – Pencil and paper based methods

Generally, risk metrics can be categorized in the following groups: pencil and paper based methods, biomechanical analysis, and strength assessment. Pencil and paper based screening approaches are most often used by practicing ergonomists (Dempsey et al., 2005).

Table 2.1 provides an overview of some commonly used checklists and pencil and paper based tools.

Table 2.1 – An overview of six commonly used risk screening or assessment tools.

Pencil and Paper Method	Physical Hazards Considered	Analysis Level	Final Output
Rapid Upper Limb Assessment (RULA) (McAtamney and Corlett, 1993)	Repetition or static load, upper limb posture, forceful exertions	Task / Element	Single score integrated from a combination of risk factors
UAW-GM Risk Factor Checklist (Keyserling, 1993)	Repetition, body posture, forceful exertions, contact stress	Job	Single score integrated from a combination of risk factors
Strain Index (Moore and Garg, 1995)	Repetition, body posture, forceful exertions	Task, combined into job	Single score integrated from a combination of risk factors
OCRA (Occhipinti, 1998)	Repetition, body posture, forceful exertions	Task, combined into job	Single score integrated from a combination of risk factors
Hand Activity Level (HAL) (Latko, 1997)	Repetition, hand posture, hand force	Job	Continuous variable for each risk factor (can be compared to the ACGIH TLV)
Quick Exposure Checklist (QEC) (David et al., 2008)	Repetition, body posture, force, vibration, stress	Task / Element	Single score for each exposure factor ranked by exposure level
Posture Activity Tools and Handling (PATH) (Buchholz et al., 1996)	Posture, Work activities, tools used, handling of tools / objects	Task / Element	Customizable

* Adapted from Ebersole, 2005

Each method provides a quick indication of the potential risk factors a worker may be exposed to during the performance of their job, or a task within their job. The primary function of each tool is to provide a means for MSD surveillance in the workplace (Silverstein et al., 1997).

They do not necessarily correspond to specific internal doses or capacities, or reveal any information about safe or appropriate exposures, though some association is implied.

2.4.2 – Biomechanical analysis

Biomechanical analysis provides more detailed information about internal doses corresponding to a given workplace exposure. Effective risk assessment through biomechanical analysis is performed by modeling internal doses and comparing those estimates to known or estimated capacity thresholds. With reference to the low back, researchers have developed methods to model internal spine compression and shear doses (e.g. McGill and Norman, 1987; Marras and Granata, 1997) that can be directly compared to known threshold limit values for spine compression (NIOSH, 1981, Jager and Luttmann, 1992) or shear (McGill et al., 1998) indicating the level of risk for a specific MSD. This process is more targeted than posture based approaches. Conversely, it requires a means to model internal tissue loading, threshold exposure levels for tissues loading, and higher fidelity information to drive model predictions, which can be costly (Winkel and Mathiassen, 1994).

Epidemiological support also underscores the importance of using biomechanical analysis in the workplace. Marras and colleagues (1995) found that five trunk motion characteristics described the relationship with risk of reporting an incidence of low back pain (odds ratio 10.7), where three of those are determined using biomechanical analysis: load moment, trunk lateral velocity and twisting velocity. Additional epidemiological support also demonstrates relationships between peak and cumulative spine compression and shear loading (Kumar 1990; Norman et al., 1998), where workers in the top 25% of loading exposures were about six times more likely to report low back pain than those in the bottom 25% of loading exposure.

Using shoulder related MSDs as an example; biomechanical analysis has a more restrictive ability to inform MSD prevention efforts in its current state. Shoulder problems are becoming an increasing concern in workplaces (Marras et al., 2009). However, few models

exist to predict shoulder internal loading in the context of ergonomic applications (an exception is Dickerson et al., 2007). Therefore, at the outset we are currently restricted in our ability to readily predict internal shoulder demands in the workplace. Research effort continues in this regard with the development of robust shoulder modeling approaches to predict internal loading (Dickerson et al., 2007); however we still lack benchmarks with which to compare this information to determine the level of risk. Unlike research on the spine, where internal loading such as compression is directly linked to mechanical bone failure (Brinckmann et al., 1988); internal shoulder loading is not specifically linked to a threshold level of mechanical exposure linked to failure. Though the ratio of compression and shear loading is important to prevent dislocation of the joint (Lippett et al., 1993), it is more likely that shoulder MSDs occur in soft tissue, as a result of complex interactions between loading, postural, kinematic and muscle changes (Michener et al., 2003). Therefore the use of biomechanical analysis in the workplace as a means to prevent shoulder related MSDs remains restricted as we continue building an effective knowledge base.

Biomechanical analysis plays a key role in explaining the relationships between external work exposures, internal doses, and their corresponding affects on tissue capacities. One key difference between biomechanical analysis and the checklists and postural tools described above is highlighted in context of the question posed by Wells (2009): “how good are our MSD risk factors?” Broadly, risk factors provide a means to screen for increased MSD risk (Silverstein et al., 1997). Checklists and posture based tools encapsulate this broad knowledge based on the strong associations uncovered between risk factors and MSDs. However, more detailed information about causal relationships between the internal dose, response(s) and capacity changes are needed to help better target intervention efforts.

Biomechanical analysis can provide this level of information. Wells (2009) demonstrates this point by highlighting that low back pain, at a biomechanical level, could be due to cumulative damage to a structure, a loss of spine stability, or “a statistically rare loading outlier in a normally innocuous task”, where each of these mechanisms could be impacted by different internal dose exposures. Therefore a broadly classified risk factor of “forceful exertion” may not provide enough information to characterize the internal biomechanical doses with respect to specific injury pathways. Further, this lack of detailed information might impact the ability to adequately inform intervention. There is a clear need for biomechanical analysis in the workplace as a means to prevent MSDs. Unfortunately biomechanical analysis, as a metric for assessing MSD risks in the workplace, remains restricted in part due to the detailed information required, and the limited data available demonstrating threshold dose levels that are problematic (Garg and Kapellusch, 2009).

2.4.3 – Strength assessment

Strength assessment is a method of predicting whether a person is capable of performing a physically demanding task without incurring an injury (Chaffin, 1975). Workers exposed to demands that are at a higher percentage of their strength capacity also have a higher incidence of injury (Chaffin and Park, 1973; Keyserling et al., 1980; Kilbom, 1988). In the context of the MSD dose-response model (Armstrong et al., 1993) illustrated above in figure 2.1, an incidence of MSD is more likely when strength dose approaches or exceeds the strength capacity, resulting in overexertion (Kumar, 2001).

Strength testing assists MSD prevention. It does so in three ways: as a tool in post-offer, pre-placement screening (Harbin and Olsen, 2005), proactive job design (Chaffin, 1997), and reactive job design (Haslegrave et al., 1997; Mital and Kumar, 1998). This type of testing

is more targeted than postural assessments because it depends on actual worker capability, as opposed to a general association between a risk factor and an MSD. However, unlike biomechanical analysis, strength testing does not provide information on tissue specific doses and responses. Figure 2.2 presents a modified version of the model presented by Armstrong et al. (1993) to demonstrate how both strength assessment and biomechanical analysis provide different types of information within this context.

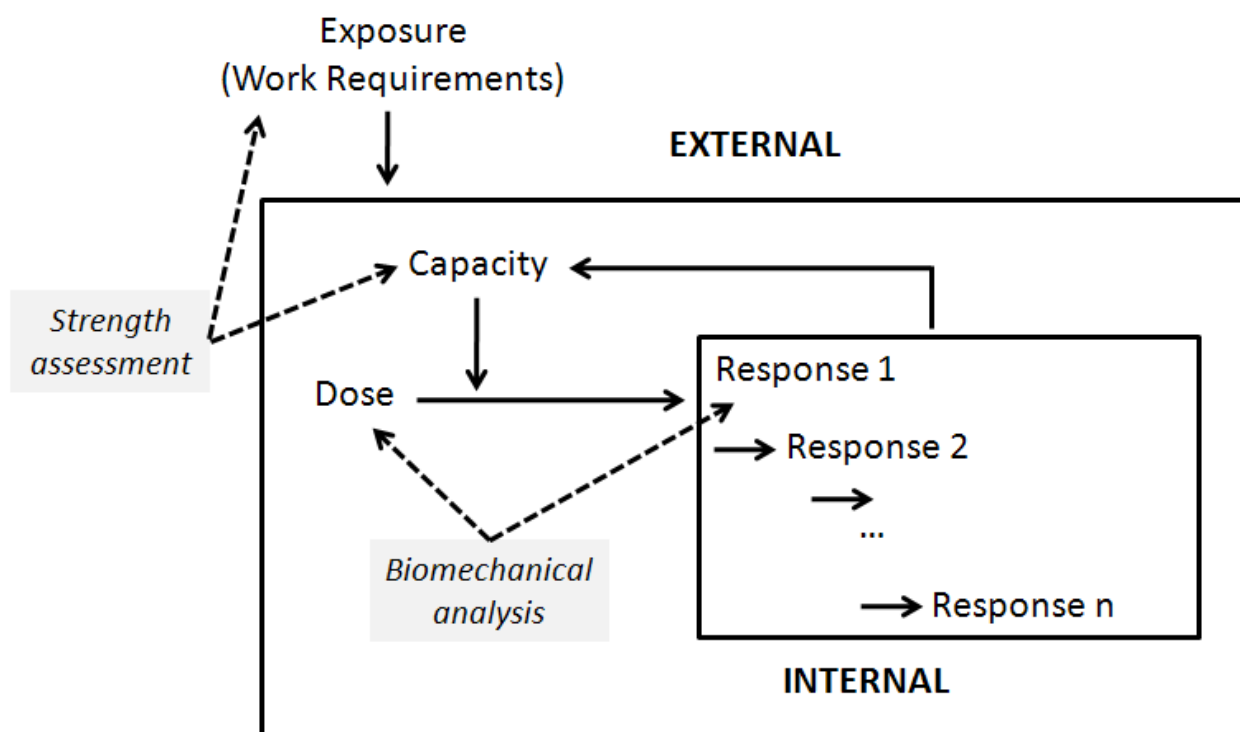


Figure 2.2 – The conceptual dose-response model (Armstrong et al., 1993) modified to demonstrate how biomechanical analysis and strength assessment can yield important information in the context of this process.

2.4.4 – The intersection of biomechanical analysis and strength assessment

Both strength assessment and biomechanical analysis provide information in the context of the dose-response model presented by Armstrong et al (1993). The critical difference is that biomechanical analysis typically deals with dose-response-capacity interactions at a tissue specific level (i.e. spine compression – fatigue fracture – reduced

capacity); whereas strength assessment deals with capacity to handle the work requirements, at the broader system level independent of what specific tissue level capacity may be limiting (i.e. the total system has capacity to lift 23 kg). Through decades of research Chaffin and colleagues (summarized in Chaffin 1997) have demonstrated that these two types of information can be combined to provide greater insight and application as a metric to prevent MSDs.

The Michigan 3-Dimensional Static Strength Prediction Program (3DSSPP) unites strength assessment and biomechanical analysis for MSD risk assessment to inform primary and secondary injury prevention. The biomechanical analysis portion of the program models joint net moment and force doses on the body based on exposure to a given work requirement. Further, the biomechanical analysis extends to the low back, where a detailed approach is used to generate estimates of spine shear and compression loading (Chaffin, 1997). Additionally, based on the joint moments calculated from the biomechanical model, a strength assessment is achieved by comparing the resulting joint moment exposures, to the population strength capacity to withstand those exposures. Collectively, this approach informs the user of a) the % of population that is likely to have adequate joint strength to perform the work requirement (strength assessment), b) the dose of compression or shear imposed on the spine due to that work requirement (biomechanical analysis), and c) the additional details regarding whole-body balance and friction requirements, and modeled muscle demands (biomechanical analysis).

In the context of the dose-response model (Armstrong et al., 1993) the 3DSSPP approach provides a link between strength assessment and biomechanical analysis. Within the software, strength capacity is dictated based on internal joint moment generating capacities, providing a direct link between strength capacity and the underlying biomechanical factor

governing that capacity. Additionally, Kerk et al., (1994; 1998) further exploit this link to estimate whole body strength capabilities based on the underlying biomechanical factors governing that capability. This combined approach provides strength related information which can be used to effectively inform both proactive job design, and reactive job design; but also, detailed biomechanical information is included, such as joint moments and forces, which can be further used to model tissue specific loading (McGill, 1996). Lastly, this approach provides a biomechanical rationale to support strength assessment by highlighting the connection between overall strength capacities and underlying biomechanical doses.

Advancing the relationship between strength capacities and underlying biomechanical features is a main theme in this dissertation. The extensive work of Chaffin and colleagues has provided a strong foundation with respect to maximum strength, or simulated job static strength using the definitions provided in Chapter 1. However, advancing this approach to include biomechanical explanations for sub-maximal (psychophysical) strength capacities is needed to continue to refine MSD prevention efforts.

2.5 – The use of psychophysical strength based metrics

2.5.1 – The evolution of psychophysics in ergonomics

Psychophysics is a branch of psychology used to investigate the relationships between physical stimuli and the sensations they produce in the body. The paradigm originated from the work of Gustav Fechner (1860), who first discovered logarithmic relationships between the magnitude of physical stimuli and the perceived sensations to those stimuli. One hundred years after Fechner's foundational research, Stanley Stevens derived the Stevens power law (Stevens, 1957) which helped revive Fechner's earlier work only this time using a power function to relate the magnitude of a stimulus (S) to the magnitude of a resulting sensation (Ψ)

in concert with a proportionality constant (k) and an exponent (n), both specific to the applied stimulus. Through a number of experiments, Stevens established the power law by measuring the sensory responses to a variety of physical stimuli.

$$\Psi = kS^n \quad (\text{eq. 2.1})$$

The power law effectively described stimulus-response relationships between sounds and loudness and light and brightness, among many other factors (Stevens, 1970). It was from Stevens' work that Snook and Irvine (1967) were able to evolve the use of psychophysics into a method for determining subjective acceptable exposure limits for workplace design.

The psychophysical paradigm originated by measuring the sensory response for a given stimulus. Snook and Irvine (1967) were less interested in quantifying the sensations that resulted from a given stimulus; rather they wanted to measure the stimulus that resulted in a given sensory response. Where Stevens provided stimuli and measured the resulting responses, Snook and Irvine asked participants to self-adjust the stimulus (weight of box lifted) to achieve a specific response ("the maximum amount you can lift comfortably without straining yourself"). The resulting box weight was then considered the acceptable limit for the lift. This initial research by Snook and Irvine (1967) transformed the psychophysical paradigm into a useful method for developing acceptable guidelines for work. Snook continued to use and evolve this approach, which eventually resulted in the development of the Liberty Mutual Manual Materials Handling Guide (©Liberty Mutual), one of the most commonly used observational techniques by certified ergonomists (Dempsey et al., 2005).

2.5.2 – Applications of psychophysics in ergonomics – lifting, pushing, pulling

Psychophysical was originally adapted for ergonomics applications focused on determining acceptable limits for lifting. Beginning with the work of Snook and Irvine (1967)

several studies were conducted to investigate how various parameters of a lift influenced the acceptable limit. The Liberty Mutual Manual Materials Handling Guide, for example, was derived from studies looking at lifting, pushing and pulling: at different frequencies, to and from different heights, using different boxes and using different handle types, etc (Table 2.2). Many other researchers have conducted similar research, but the efforts of Dr. Snook at Liberty Mutual represent the most concentrated thrust using psychophysical estimation to determine load acceptability.

Table 2.2 – List of published psychophysical research performed at Liberty Mutual used to develop the Manual Materials Handling Guidelines.

Authors	Year	Title
Snook & Irvine	1967	Maximum acceptable weight of lift
Snook & Irvine	1969	Psychophysical studies of physiological fatigue
Snook et al.	1970	Maximum weights & workloads acceptable to male industrial workers
Snook	1978	Ergonomics Society Lecture: The design of manual materials handling tasks
Ciriello & Snook	1983	A study of size, distance, height, and frequency effects on manual handling tasks
Ciriello et al.	1990	The effects of task duration on psychophysically determined maximum acceptable load changes
Snook & Ciriello	1991	The design of manual materials handling tasks: revised tables of maximum acceptable weights and forces
Ciriello et al.	1993	Further studies of psychophysically determined maximum acceptable weights and forces
Ciriello et al.	1999	Maximum acceptable forces of dynamic pushing, a comparison of two techniques
Ciriello	2001	The effects of box size, vertical distance, and height on lowering tasks
Ciriello et al.	2007	Revisited: Comparison of two techniques to establish maximum acceptable forces of a dynamic pushing task

In addition to providing limits for lifting, pushing and pulling, the psychophysical approach has also been applied to investigate how work parameters influence the perception of acceptability. With respect to psychophysically acceptable lifting, researchers have examined the role of lifting style (Garg and Saxena, 1979), asymmetry of lift (Garg and Banaag, 1988; Han et al., 2005), the role of footwear and floor conditions (Li et al., 2007; Wickel and Reiser, 2008) and the effects of obesity (Singh et al., 2009). These investigations continue to inform the understanding of how to (re-)design work to maximize work productivity by creating parameters that result in the highest acceptable limits.

2.5.3 – Applications of psychophysics in ergonomics – upper extremity motions

Psychophysical force and moment estimation has also been used to examine acceptable limits for the upper limb. Several additional applications for psychophysics beyond force and moment estimation have also been used to determine “acceptability” criteria (e.g. Sood et al., 2007); however this work will focus on the force or moment estimation application of psychophysics. Similar to the psychophysically acceptable lifting research, the literature on upper limb specific psychophysical limits again includes work from Snook and colleagues where they examined psychophysical limits for a number of hand motions and gripping tasks (Snook et al., 1995; 1997; 1999; Ciriello et al., 2001; 2002). One key difference that accompanied their transition into upper limb psychophysical limits was a switch from using task based definitions (lifting, lowering, etc.) to joint motion based definitions (wrist flexion, ulnar deviation, etc.). In the context of the global thesis being investigated here, this difference is important. By testing psychophysically acceptable moments for joint specific motions, the acceptable moment is inherently related to the strength capacity for that motion (i.e. the psychophysically acceptable force cannot be more than the maximum moment capability). This provides support to the hypothesis that psychophysical limits are at least in part related to maximum moment capacity.

Upper limb psychophysical limits have also been derived using task based definitions. Though other researchers have contributed to the upper limb psychophysical literature, two research groups have been prolific in their contributions in this regard. Jeffery Fernandez’s group has examined gripping, pinching, drilling, and riveting (Kim and Fernandez, 1993; Dahalan and Fernandez, 1994; Davis and Fernandez, 1994; Marley and Fernandez, 1995; Klein and Fernandez, 1997; Fredericks and Fernandez, 1999). Jim Potvin and colleagues have

examined hose insertions, fastener initiations, pinches, and hand impacts (Potvin et al., 2000; Cort et al., 2006; Potvin et al., 2006; Andrews et al., 2008).

In addition to providing a variety of task based threshold limits, each group has also made a unique contribution. Fernandez demonstrated how the traditional load-adjust approach to psychophysics - where the participant self selects the load - could be altered to a frequency-adjust approach - where the load is constant but the participant self-selects the work rate. This contribution does not directly impact the work in this dissertation. However, it does provide evidence that frequency (exposure time) of motion is also perceivable. Therefore it is important to note that the relationships investigated through this dissertation are dependent on the exposure time as well. Potvin and colleagues contributed the idea that psychophysical force is proportionally related to maximum voluntary force (Potvin, 2007). Although this idea has been suggested by others, (Nussbaum and Lang, 2005), Potvin demonstrated it quantitatively (Potvin, 2007). This dissertation aimed to evaluate this proportionality relationship and identify if it was supportable through a biomechanical explanation.

Two other notable contributions further our understanding of upper limb psychophysical limits. Moore and Wells (2005) extended the work addressing the effect of frequency on psychophysical limits. Their research demonstrated that duty cycle, rather than cycle time (frequency), most substantially affected psychophysically acceptable forces. In the context of the weakest link approach described in this dissertation, this finding is important. If the psychophysically acceptable force is directly related to the joint moment capacity (during strength limited exertions) then it could be expected that the acceptable forces would decrease as the time required to sustain the relative moment demands increased (Rohmert, 1973).

The second notable contribution supports the hypothesis that psychophysically acceptable forces for the upper limb are additionally dependant on time related factors. Nussbaum and Johnson (2002) modeled psychophysically acceptable finger forces using the following equation:

$$K = F^{0.5} \times MAL(\%)^2 \quad (\text{eq. 2.2})$$

In their equation, $MAF\%$ was the psychophysically acceptable force as a percentage of the maximum voluntary force, K was a finger specific constant, and F referred to the frequency (exertions / min). The key point illustrated by the equation is that psychophysically acceptable forces are likely related to the maximum acceptable force and that this relationship is scalable based on a time sensitive factor. This underscores the importance of testing the relationship between psychophysically acceptable forces and maximum voluntary forces in conditions where the influence of a time sensitive factor is minimal and controlled.

2.5.4 – Advantages and disadvantages of using psychophysical metrics for setting threshold exposure limits in the workplace

Psychophysical strength has many advantages with respect to setting exposure thresholds in the workplace. The following list of advantages has been reported in the literature (Ayoub and Dempsey, 1999; Snook, 1999; Dempsey, 2006):

1. Psychophysics allows for the realistic simulation of industrial tasks.
2. Psychophysical results are consistent with the industrial engineering concept of "A fair days work for a fair days pay".
3. Psychophysical results are very reproducible.
4. Psychophysical judgments take into account the whole job, integrating biomechanical and physiological factors.

5. The psychophysical approach is less costly and less time consuming to apply in industry than many of the biomechanical or physiological techniques.
6. Currently, psychophysical data represent one of the few quantitative guidelines for the design of force limits in upper extremity intensive tasks.

Additionally, the following limitations and disadvantages were also described:

1. Psychophysics is a subjective method by definition.
2. The assumption that the subjective workloads selected by participants are below the thresholds for injury has not been validated. There is no epidemiological support for using psychophysical data in the design of upper extremity extensive tasks to prevent injury.
3. The range of data for designing upper extremity tasks is somewhat limited at this time.

The scope of this dissertation aims to provide information that will help address these three limitations by investigating a possible relationship between psychophysical strength capacity, simulated job static strength capacity and the underlying biomechanical doses.

Psychophysical strength provides important information despite it being viewed as a subjective measure. Biomechanical analysis and strength assessment each provide useful quantitative information (section 2.4.4). As 3DSSPP demonstrates, a link between strength capacity at a whole body level and internal biomechanical doses and capacities at an internal level (Figure 2.2) can yield additional knowledge. Considering that psychophysical strength provides additional information that is different from biomechanical and physiological assessment (Garg and Ayoub, 1980; Nicholson 1989; Dempsey, 1998; Keyserling et al., 2000) it is useful to investigate how psychophysical strength may also be related to both

biomechanical and maximum strength capacity information. Within the context of the dose-response model (Armstrong et al., 1993), psychophysical strength provides insight into how much exposure can be withstood without causing responses that generate feelings of pain or discomfort (Snook et al., 1995) (Figure 2.3). By linking psychophysical load selection with underlying biomechanical doses and maximum strength capacities, we aim to address the limitation that psychophysical limits are solely subjective, and further improve our ability to predict these types of limits in the workplace

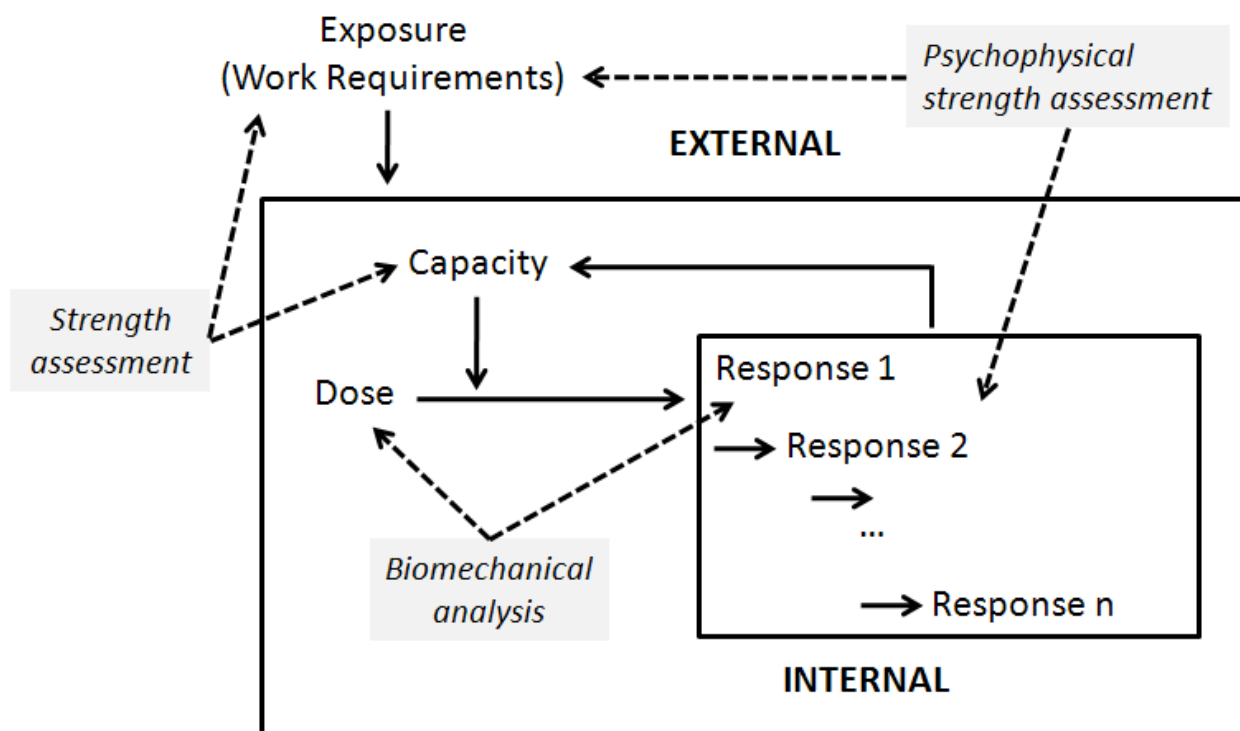


Figure 2.3 – The conceptual dose-response model (Armstrong et al., 1993) modified to demonstrate how biomechanical analysis, strength assessment and psychophysical assessment can yield important information in the context of this process.

Where maximum strength assessment reveals the maximum system capacity, and biomechanical analysis reveals specific internal doses and responses, psychophysical assessment reveals information about what work exposure is acceptable without causing responses related to uncomfortable sensations. By expressing these features overlaid on a dose

response model, it is inherently implied that they all relate to injury. However, as noted in the list of limitations, psychophysically acceptable strength thresholds may not necessarily protect against injury development (Ayoub and Dempsey, 1999; Snook, 1999). Nonetheless, by viewing each of these factors in this context it also implies that there is a link between them. This thesis is directed at examining the relationships between maximum strength, psychophysical strength and underlying biomechanics. And although the goal is not to demonstrate that working at or below psychophysically selected exposures protects against injury, the objective of this work is to demonstrate if a quantitative relationship between these factors exists.

2.5.5 – The link between biomechanics and psychophysics

This dissertation is not the first attempt to examine a biomechanical explanation for psychophysical load selection. Explanations based on electromyographic findings have been discussed most frequently in the literature. Table 2.3 summarizes those findings.

Table 2.3 – A summary of EMG findings reported in the literature when conducting psychophysical research on upper extremity intensive tasks.

Reference	Muscles	EMG Processing	Results
Klein & Fernandez, 1997	FCR, ECR	<ul style="list-style-type: none"> •Calculated RMS and Median Frequency •Sampled at 1024 Hz 	RMS increased and MF decreased with increases in wrist flexion angle, pinch force, or duration
Marley & Fernandez, 1995	Flexor carpi ulnaris, Anterior deltoid	<ul style="list-style-type: none"> •Calculated RMS •Sampled at 800 Hz 	Increased RMS with increased wrist flexion (FCU) or increased shoulder flexion (Delt)
Davis & Fernandez, 1994	Forearm flexor and extensor muscles	no details reported	no results reported
Kim and Fernandez, 1993	Forearm flexor and Adelt	<ul style="list-style-type: none"> •Calculated RMS •Sampled at 512 Hz 	Increased RMS with increased force requirements
Fredericks and Fernandez, 1999	Forearm flexor and extensor muscles	<ul style="list-style-type: none"> •Calculated RMS and MF (note EMG electrodes repositioned for every measurement) •Sampled at 800 Hz 	Increased RMS in flexors with increase in wrist flexion
Dahalan and Fernandez, 1993	FD, ECR	<ul style="list-style-type: none"> •Calculated RMS and Median Frequency •Sampled at 1024 Hz 	RMS increased and MDF increased with increases in contraction intensity
Cort et al., 2006	ECU, FCU, BB, BR, FDI, TNR	<ul style="list-style-type: none"> •Normalized EMG to MVC •Sampled at 1000 Hz 	Low activity as a %MVC, phasic response of the thenar muscle suggests its importance in fastener initiations
McFall, 2008	ECU, ED, ECR, FCU, FCR, FDS, FPL, FDI	<ul style="list-style-type: none"> •Normalized EMG to MVC obtained during pinch and power grips •Calculated APDF's and MPF •Sampled at 1000 Hz 	No decrease in MPF over the course of individual trials Peak %MVC EMG fell below recommended static (10th%ile) and dynamic (50th%ile) recommended thresholds
Moore, 1999	FDS, ECRB, FCR	<ul style="list-style-type: none"> •Normalized EMG to initial measurement of the day •Calculated average %change in EMG, MF and MPF •Sampled at 1000 Hz 	Average EMG for FCR increased from the start to end of the day

* FCR – flexor carpi radialis, ECR – extensor carpi radialis, Adelt – anterior deltoid, FD – flexor digitorum, ED – extensor digitorum, ECU – extensor carpi ulnaris, FCU - flexor carpi ulnaris, BB – biceps brachii, BR – brachioradialis, FDI – first dorsal interosseous, TNR – thenar, FDS – flexor digitorum superficialis, FPL – flexor palmaris longus, ECRB – extensor carpi radialis brevis.

This body of work demonstrates that EMG changes occur as task parameters change; however, acceptable forces and frequencies are chosen by workers such that the corresponding EMG amplitudes remain below identified threshold levels (Jonsson, 1978), and further EMG mean power frequency remains consistent. These results are consistent with Keyserling et al. (2000) suggesting that psychophysical estimation is sensitive to fatigue. Though insightful, the EMG results alone do not provide the level of information needed to more adequately address the plausibility of a biomechanical explanation for psychophysical force selection.

To date, only one study has calculated the joint moment demands associated with psychophysical load selection (Nussbaum and Lang, 2005). Corresponding relative joint demands were calculated during each psychophysical load selection by dividing the joint moment (L4/L5, shoulder, elbow) by the maximum joint strength (moment) calculated based on strength tests. Psychophysically acceptable loads were selected such that the highest relative joint demand experienced by any single joint was equal to approximately 70% of the maximum strength at the joint that was biomechanically most limiting. The kinematic and kinetic results presented by Nussbaum and Lang (2005) lays the foundation for the current work to further investigate the relationship between joint demands and psychophysical force selection hinging on an understanding of the capacity at the weakest or limiting joint or factor.

2.6 – Methodological considerations for assessing psychophysical strength

Many researchers have scrutinized aspects of the traditional psychophysical protocol. Those aspects investigated most often include: participant selection, training, psychophysical instructions, and the data recording length.

2.6.1 – Participant selection

Participant recruitment is an important concern in the development of a research experiment. Three specific participant related concerns have been addressed in the literature with respect to their influence on psychophysically acceptable forces. Work experience is considered to be the most critical of these concerns. Skilled workers, familiar with the task respond differently than unskilled workers not familiar with the task (Gamberale, 1987; 1988; Potvin et al., 2000). Further, inexperienced students selected psychophysically acceptable forces at significantly higher levels than industrial workers (Johnson and Nussbaum, 2003). These findings support the conviction that industrial workers should be used when determining psychophysical guidelines for industrial work because “students, housewives and military personnel may have different perceptions of industrial work” (Snook, 1978). This dissertation is directed towards having an applied application in the long term. Therefore in light of the known influence that work experience can have on estimates of psychophysical force, workers familiar with general manual materials handling were hired as participants.

Gender and age may also impact the magnitude of psychophysical estimates. Indeed, females do tend to select psychophysical forces at a lower magnitude; however, when this difference is considered, men and women respond similarly (Ciriello and Snook, 1983; Ciriello et al., 1990; Nussbaum and Johnson, 2002; Wickel and Reiser, 2008; Singh et al., 2009). In terms of age, Wright and Mital (1999) reported no considerable difference in psychophysically acceptable loads between people aged 55 - 74 and 18-35. Using an age blocked design; both Cort et al. (2006) and Potvin et al. (2006) corroborated this finding with reference to fastener initiation frequencies, or grasp, pinch, and finger press forces. Collectively these results

indicate that psychophysically acceptable forces will likely be similar across workers stratified by age; however, females are likely to select forces at a lower level than males.

2.6.2 – *Training*

The determination of psychophysically acceptable loads requires participants to make a choice about a workload that they feel would be acceptable for them to perform for a defined number of repetitions over the course of a work shift. To ensure that participants are able to make this choice and understand how different magnitudes of workload can affect their feelings of wellness, it has become standard operating procedure to provide a training or familiarization period prior to psychophysical testing. However some controversy exists regarding the length or type of training required. Training prior to psychophysical experiments has varied from 4-5 days (Snook et al., 1995; 1997; 1999) to 2 hours of hands on training (Kim and Fernandez, 1993), to watching a video for familiarization (Nussbaum and Johnson, 2002). Though research demonstrates that unskilled workers can be trained to produce psychophysical forces that converge with those of skilled workers (Potvin et al., 2000) no conclusive research has clearly demonstrated how much training is required for a skilled worker to become familiar.

2.6.3 – *Psychophysics instructions*

Within the paradigm of psychophysics, participants are asked to select a parameter (in this case force) such that their perception to the response of selecting that parameter is below some criteria. For example, when asking workers to select an acceptable weight for lifting Snook and Irvine (1967) told workers “We are not interested in the maximum amount of weight that can be lifted, but only in the amount that can be lifted *comfortably* and without *strain*”. This statement provided the participants with the knowledge that they were to gauge

their lift weight according to their perception of their *comfort* level and *strain* level. The specific content of this verbal message can have an impact on psychophysically acceptable forces (Karwowski et al., 1999). In addition, the frequency at which the instructions are provide can also impact estimates (Gamberale, 1987; 1988). Therefore it is prudent to maintain a consistent form of instruction, provided at regularly occurring intervals to ensure that these factors do not adversely affect psychophysical estimates between or within participants.

2.6.4 – Time required to obtain a stable psychophysical estimate

There is no consensus on the appropriate length of time over which to determine psychophysically acceptable loads. Nussbaum and Johnson (2002) found that estimates of acceptable forces for single digit pushes stabilized quickly (within 5 minutes), and did not significantly vary over the remainder of the 25 minute load-adjust protocol. Ciriello et al. (1990) report no significant change in lifting or lowering loads over the course of a four hour protocol. Conversely Snook et al. (1995; 1997; 1999) reported a small, but significant decrease in acceptable torque levels from the first hour, until the fifth hour, where the acceptable torques leveled off for the remainder of the seven hour collection. In addition, Fernandez et al. (1991) demonstrated that load decreased over the course of an 8-hour work shift, although most changes occurred in the first hour (46% of all adjustments), or fifth hour, immediately following the allotted lunch break (24% of all adjustments). The lack of a clear timeline to obtain stable estimates is troubling. However, in a practical sense, this indecision in length of time is rationalized based on the high cost to perform the research and the small difference in magnitude that would be expected to occur by using longer lengths (Snook et al., 1995; 1997; 1999).

2.6.5 – General conclusions on choosing a psychophysical protocol

The outcomes of this dissertation hinge on being able to adequately determine psychophysically acceptable forces. Throughout section 2.6, a number of considerations have been presented with reference to their impact on psychophysically acceptable forces. It is clear that some factors (skilled vs. unskilled workers) appear to have more of an affect than others (age). Each of these considerations and factors were considered in the design of the psychophysical studies performed as part of this dissertation. The purpose of this section was to provide the rationale to support those study designs.

Chapter 3 – The roles of whole body balance, shoe-floor friction and joint strength during maximum exertions: Searching for the “weakest link”.

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Submitted to Applied Biomechanics

3.1 – Overview

Exerting manual forces is critical during occupational performance. Considering how specific factors influence performance capacity, including balance, friction, or joint strength could help discern situational underlying sources of hand force capacity. This research focused on identifying how these factors limited hand force capability during unilateral pulling, pressing down, and medial exertions. These efforts were performed in a self-selected manner under four conditions in which different body regions were constrained, thereby removing potential limiting factors. Centre of pressure movements (COP), upper body joint strengths, and muscle activation were monitored.

Joint strength limited downward forces – specifically, shoulder, trunk or potentially lower body strength. Whole body balance limited pulling, with maximal force occurring when the COP excursions reached a functional limit within the base of support (BOS). While braced at the trunk and pulling, COP excursions surpassed BOS limits by 400% while hand force increased nearly 300%. Medial exertion strength was modified by balance, friction and joint strength; but a clear limiting factor did not emerge. Medial force capacity may be limited by trunk strength, but is also influenced by both friction and balance. Depending on the specific circumstances, balance, friction, and joint strength appear to differentially limit the ability to exert manual forces.

3.2 – Introduction

Incorporating human force producing capability into job design can mitigate workers' risk of injury. Since the nineteenth century, researchers have measured force production in an effort to match individual capability with anticipated performance demands (Sargent, 1887) to optimize performance and minimize injury risk. The relevance of this concept persists, as the risk of injury increases when working at or above capability (Chaffin et al., 1978). However, measuring individual worker capabilities to facilitate this matching is time and cost intensive. Alternatively, models designed to predict force producing capability could enable designers to match job demands with prospective worker capabilities. However, there have been few attempts to develop comprehensive predictive models (Grieve 1979a; 1979b; Kerk et al., 1994), and none designed to incorporate three dimensional tasks. This may be due to an incomplete understanding about how force capacity is limited.

Several factors, both extrinsic and intrinsic to the worker, can limit force producing capability during manual material handling exertions. Limiting factors include whole body balance (Kerk et al., 1998; Holbein and Chaffin, 1997), shoe-floor friction (Kroemer, 1974), hand-handle friction (Seo et. al., 2010), and individual joint moment strength (Chaffin, 1997). Whole body balance becomes limiting when the centre of pressure (COP) approaches the limits of support (BOS). As a worker exerts a force, the reaction force creates a moment about the workers centre of mass (COM). Assuming the worker remains in a static posture; the COP must be displaced to ensure the moment caused by the ground reaction force is equal and opposite to the moment caused by the reaction force at the hand. Gaughran and Dempster (1956) demonstrated this experimentally while measuring horizontal force capability. Hand force production was directly proportional to the length of the BOS, where higher hand forces

were recorded when the BOS was extended. Recent work investigating balance and hand force production has shown that COP excursions are further limited to an area within the BOS, the “functional base of support” (FBOS), (Holbein and Chaffin, 1997; Holbein and Redfern, 1997; Holbein-Jenny et al., 2007). Quantifying COP excursions within the FBOS may help infer how balance constrains capability. Delineating FBOS limits during maximum pulling and pressing efforts should aid future modeling approaches.

Joint strength has been identified as a fundamental limiting factor to hand force production capability. Through decades of research, Chaffin and colleagues conceptualized the use of joint strength as a constraint on capability in the workplace (Chaffin, 1969; Chaffin and Baker, 1970; Garg and Chaffin, 1975; Chaffin et al., 2006) in the development of the Michigan Three Dimensional (3D) Static Strength Prediction Program (3DSSPP). The strength constraints in the 3DSSPP are based on work by Schanne (1972) and Stobbe (1982) who developed population scalable strength estimates to predict maximum joint strength for each joint, depending on the posture and exertion type. Although these estimates of joint strength are widely cited, modeling of the actual joint strengths (moments) generated in simulated occupational activities when joint strength is known to be limiting is scarce. Moreover, comparison of measured joint strengths and estimated maximum joint strengths is rare.

Shoe-floor friction and hand-handle friction can also be limiting factors, though they may be less likely to limit performance under most conditions. Grieve (1983) concluded that friction is most likely the limiting factor only during conditions where the coefficient of friction is greatly reduced. Similarly, Seo et al., (2010) demonstrated that hand-handle friction is also most limiting only when it is greatly reduced, and further, it can be eliminated as a constraint if the handle is perpendicular to the hand.

The purpose of this research was to identify when specific factors (whole-body balance, shoe-floor friction, and joint strength) limited the ability of the distal upper limb to exert forces on the environment during unilateral pulling, pressing down, and medial exertions. Further, we quantified maximum COP excursion, maximum joint strengths, and maximum muscle activation achieved in the context of these factors. We aimed to provide improved clarification for the influence of limiting factors on hand force production.

3.3 – Methods

3.3.1 – Participants

A convenience sample of eighteen right hand dominant male university students (stature: 1.79 ± 0.08 m; body mass: 80.5 ± 10.2 kg) participated. According to an a priori power analysis, a minimum of ten participants was necessary to provide >80% power to detect significant differences in hand force between conditions with an effect size greater than 1.0 (partial η^2 greater than 0.5 when using a repeated measures design) (Faul et al., 2007; 2009). The study exclusion criteria were self-reported upper extremity or low back disorders or pain within the previous year. The research protocol was approved by the university research ethics review board.

3.3.2 – Instrumentation

Ten muscles were monitored using surface electromyography (sEMG). Bi-polar silver silver-chloride Noraxon dual surface electrodes with a fixed 20-mm inter-electrode spacing (Noraxon, Arizona, USA) were placed over each muscle belly (Table 1). Prior to electrode placement, the skin overlaying the muscle was shaved and cleaned with alcohol to minimize impedance. EMG signals were acquired using the Noraxon Telemetry 2400 T G2 telemetered EMG system (Noraxon, Arizona, USA) and A/D converted at 1500 samples/second using a

16-bit A/D card with a $\pm 3.5V$ range (VICON, Oxford, UK). The recording system included band pass filtering (10-500 Hz) and differential amplification (common-mode rejection ratio >100 dB at 60Hz, input impedance $100M\Omega$) of the detected signal. Force was measured using an AMTI 6 degree-of-freedom transducer (MC3A, AMTI MA, USA), rigidly fixed to a custom handle to allow the participant to exert force in the specified directions (described below). Force was sampled synchronously with sEMG at 1500 Hz using VICON Nexus 1.2 software (Oxford, UK).

Table 3.1 – A list of the muscles recorded using sEMG, the electrode locations and MVIE tests used to determine muscle specific MVEs. Surface electrodes locations and MVIE tests were adapted from SENIAM^a, McGill, 1991^b, and Delagi et al., 1980^c.

Muscle	Surface electrode location	MVIE test
Bilateral Rectus Abdominus ^b	Approximately 3 cm lateral to the umbilicus, over the muscle belly	Lying supine with trunk elevated 60 degrees from horizontal, resist against trunk flexion
Bilateral Erector Spinae - longissimus ^b	Approximately 5 cm lateral to the T9 spinous process, over the muscle belly	Lifting the trunk from a prone position
Right upper trapezius ^a	Placed at 50% of the distance on the line from the acromion to the spine on c7 vertebra, over the muscle belly	Elevate the acromial end of the clavicle and scapula. Apply pressure against the shoulder in the direction of depression and against the head in the direction of flexion anterolaterally.
Right middle trapezius ^a	Placed at 50% of the distance between the medial border of the scapula and the spine, at the level of T3, over the muscle belly	To position the scapula and to obtain leverage for the test, the elbow needs to be extended and the shoulder placed in 90 degrees abduction and lateral rotation. This rotation of the shoulder is denoted by the position of the hand with the palm facing cranially (without elevating the shoulder girdle)
Right latissimus dorsi ^b	Lateral to the T9 spinous process, over the muscle belly	Resisted lat pulldown with internal rotation of the humerus
Right teres major ^c	Three fingerbreadths above the inferior angle of the scapula along the lateral border	Resisted lat pulldown with internal rotation of the humerus
Right pectoralis major - sternal origin ^c	Over the anterior axillary fold, 6 cm above the nipple, directed superior-laterally	Press palms together with arms flexed to 90° degrees and elbow slightly bent
Right pectoralis major - clavicular origin ^c	Over the anterior axillary fold, between the sternoclavicular joint and the corocoid process, 2 cm below the clavicle, directed inferior-laterally	Press palms together with arms flexed to 90° degrees and elbow slightly bent
Reference electrode	Place over the clavicle	

Three-dimensional motion was tracked at 50 Hz using an 8-camera Vicon MX20 System (Vicon, Oxford, UK). Thirty-eight individual markers were placed over anatomical landmarks including the C7 and L5 vertebrae, over the suprasternal notch, xiphoid process, and bilaterally over the 2nd and 5th metacarpals, radial and ulnar styloids, medial and lateral epicondyles, the acromion, ear, anterior superior iliac spine, greater trochanter, medial and lateral condyles of the knee, medial and lateral malleolus, the tip of the 1st and 5th metatarsals, and at the posterior border of the calcaneus. Additional marker clusters secured on rigid plates, were positioned over the sternum, and bilaterally over the forearm, upper arm, leg, shank, and over the top of the foot. The marker clusters were used to track segment movement during experimental testing. A static calibration frame established the relationship between the clusters and the calibration markers over the anatomical landmarks, and subsequently joint centers and segment coordinate systems were described (Kingma et al., 1996).

3.3.3 – Protocol

Participants completed six different maximum voluntary isometric exertion (MVIE) tests (Table 1) under manual resistance, where each test was selected to elicit the maximum electrical activity from each of the muscles respectively. Each MVIE test was performed three times where the trial with the highest maximum voluntary electrical activity (MVE) was used to as the reference value to normalize subsequent sEMG data (Winter, 1991). Participants were given 2 minutes of rest between MVIE tests (Chaffin, 1975; Mathissen et al., 1995).

Participants were provided with approximately five minutes to warm-up and practice the MVIE tests in the postures outlined in Table 3.1. Twenty minutes of rest was provided before participants began the experimental trials.

For the experiment, participants completed maximal exertions in the medial (left across the body), horizontal pull (inwards towards the body), and downward press (towards the floor) directions against a handle using a power grip. Each exertion was performed twice within each of the five experimental conditions (described below). Participants completed each exertion within a five second window and were asked to ramp up to their maximum force during the first 1-2 seconds, and then sustain that maximum until the end of the five seconds (Chaffin, 1975). A minimum of two minutes of rest was provided in between exertions (Chaffin, 1975). The five experimental conditions were:

1. Shoulder Width Foot Posture (SWFP): Each participant stood with their feet shoulder width apart (Figure 1-A). This condition represented a generic working position.
2. Free Foot Posture (FFP): Without altering the position of the torso and shoulder relative to the handle, participants were given up to five minutes to test different foot placements to determine a position that would allow them to produce the most force in the required direction (Figure 1-B). This condition represented a generic work posture where participants could choose their own preferred posture to maximize hand force.
3. High Friction (HF): Participants stood with their feet shoulder width apart, similar to Figure 1-A with the soles of their shoes taped to the floor using double sided indoor Scotch carpet tape (3M, MN, USA) . This condition was intended to eliminate friction as a possible limiting factor.
4. Lower Body Braced (LBB): The upper legs were braced with a rigid fixture (Figure 1-C, with only the legs braced) with the feet positioned and still taped similar to the HF condition. This condition was intended to eliminate balance as a possible limiting factor.

5. Upper Body Braced (UBB): Both the lower and upper body were braced (Figure 1-C).

This condition was intended to eliminate the influence of trunk strength as a possible limiting factor, and to further reduce balance as a possible constraint.

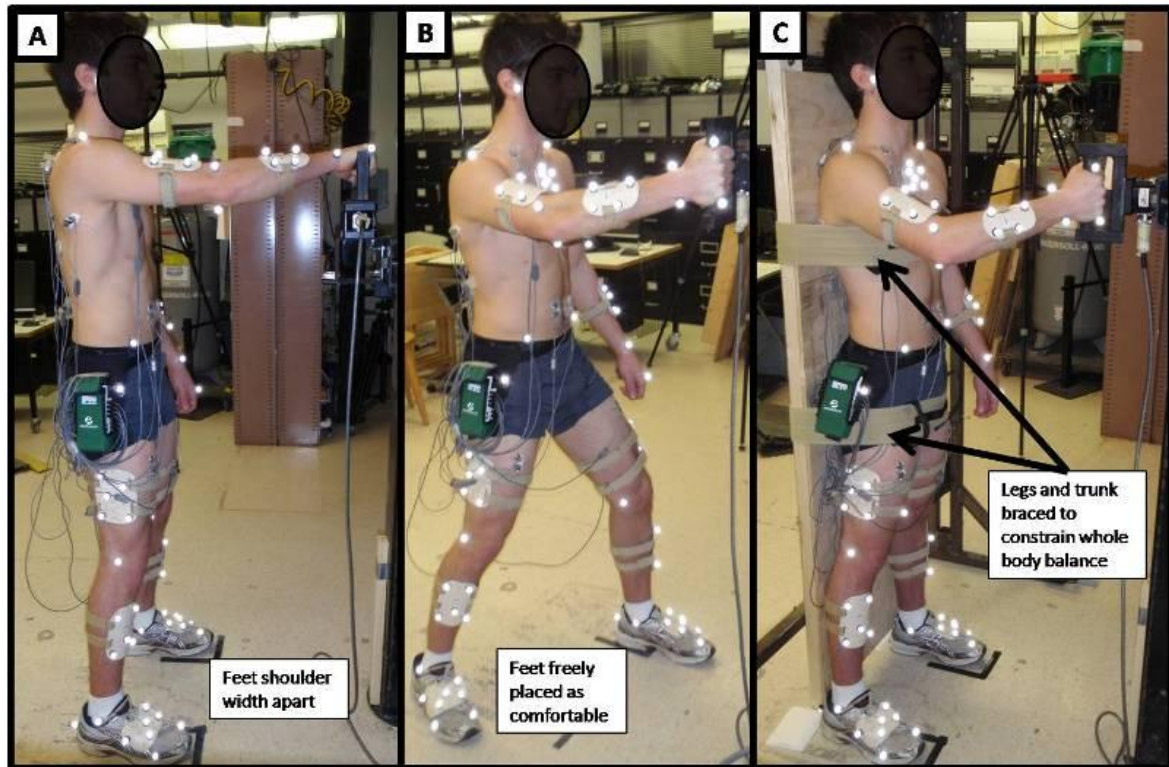


Figure 3.1 – The experimental conditions tested. Participants completed downward presses, pulls, and medial pushes in five conditions: 1) shoulder width foot placement (“SWFP” seen in frame A), 2) free foot placement (“FFP” seen in frame B), 3) high friction (“HF” seen in frame A, with feet taped to floor), 4) lower body braced (“LBB” seen in frame C, without upper body chest strap), and upper body braced (“UBB” seen in frame C).

Exposure to experimental conditions was block randomized, whereby participants completed exertions within the blocks of SWFP, FFP, and the group of HF, LBB, and UBB in a random order. The exertion direction was randomized within each block. For all exertions the handle was positioned at shoulder height, along the midline of the body, at a distance of approximately 80% of limb length for all exertions.

3.3.4 – Data analysis

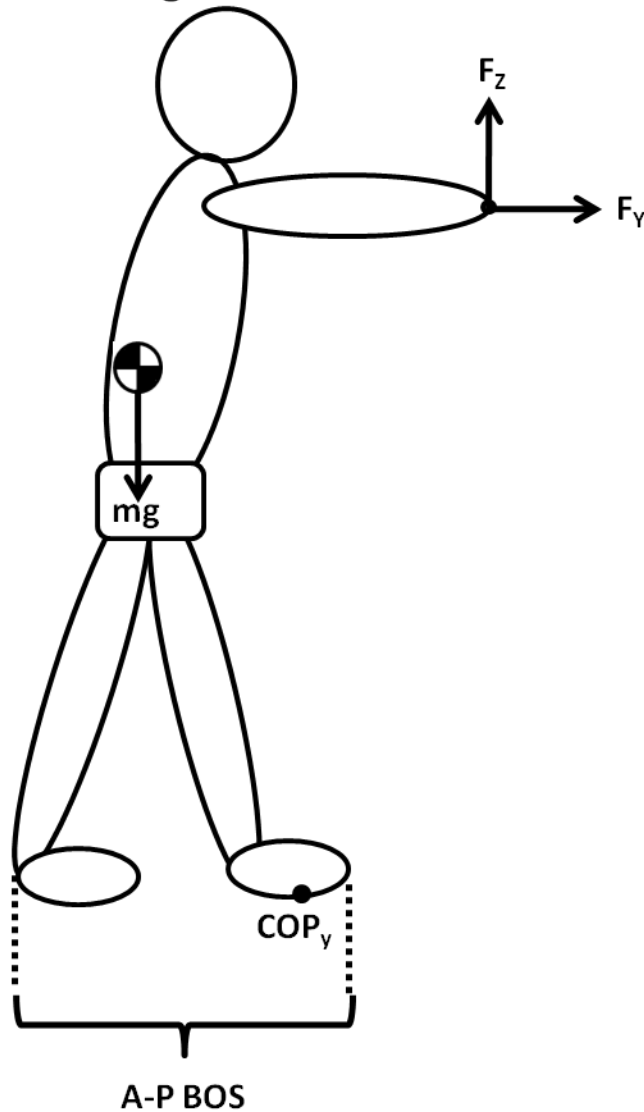
All sEMG signals were linear enveloped by full wave rectifying and digitally filtering the data (Winter, 2005) at 4 Hz using a single pass 4th order low pass Butterworth filter (Mathiassen et al., 1995). Linear enveloped sEMG was then expressed relative to the peak activity measured in the MVIE tests, yielding a time series %MVE. A single representative %MVE value was then determined for each muscle in each trial as the mean %MVE between the 2nd and 3rd second of each trial (for a detailed schematic please see Appendix 3a). The EMG was processed to determine how active each muscle was relative to the maximum amount of activity that could be generated in an optimal, standardized posture. An average of the signal between the 2nd and 3rd seconds was used as the representative amount of activity required to sustain the contraction.

Peak hand force was determined as the peak value resulting from a 500 millisecond moving window average over the raw force trace (for a detailed schematic please see Appendix 3b). The corresponding postural data was also extracted and averaged over the same 500 millisecond window. This ensured that both the force and postural data from the same instance in time were used for subsequent analysis. A low pass filter was used to remove electrically induced noise from the signal and a sliding window average was used to smooth the effects of tremor and small jerking motions applied to the handle.

Shoulder moments were calculated using a 3D static linked segment model (Dickerson et al., 2007) using the peak force and corresponding joint positions extracted from the postural data. Using the assumptions that the body is both rigid and static during the exertion, the location of the COP was solved for by summing the moments acting about the COP. The opposing moments are caused by the reaction force acting at the hand and the mass of the body

acting at the whole body center of mass (Figure 3.2). From the figure, it is clear that the A-P COP location is dependent on the components of force applied in the vertical (F_z) and along the AP horizontal axis (F_y), while the M-L COP location is again dependent on the force applied in the vertical (F_z) axis, but also along the ML horizontal axis (F_x). The vertical location of the COP is assumed to always be on the floor, and no axial rotation moments about the COP were determined. By simplifying the AP and ML COP calculation into two separate 2D analyses, the corresponding moment arms required to solve the AP and ML moment equilibrium equations could be obtained from the marker data. Once the COP location was obtained in global coordinates, the AP COP was expressed as the distance from the BOS center normalized to the length of the BOS. The ML COP was similarly expressed as the distance from the BOS centre normalized to the breadth of the BOS. The geometric centre of the BOS and the BOS dimensions were obtained from marker data (the lateral malleolus, the tip of the 1st and 5th metatarsals, and at the posterior border of the calcaneus – Holbein-Jenny et al., 2007).

Anterior – Posterior COP
Sagittal View



Medial – Lateral COP
Posterior View

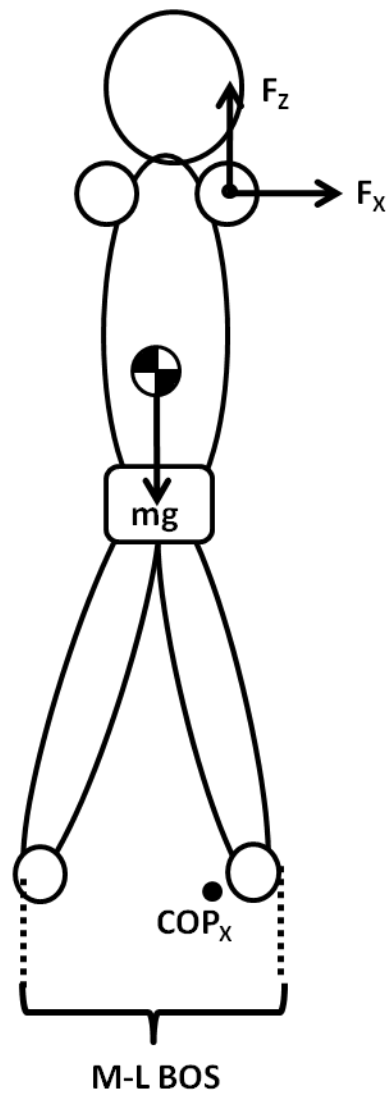


Figure 3.2 – A schematic illustration to demonstrate how the COP location was determined by solving for the moment about the COP in both the AP and ML planes. The simplification allowed COP (x,y) to be determined; however it does not allow for any axial rotational moments to be determined.

3.3.5 – Statistical analysis

A repeated measures analysis of variance (ANOVA) was used to examine the effect of condition (SWFP, FFP, HF, LBB, UBB) on the dependent variables: %MVE (for each muscle)

peak hand force, shoulder moment (about each anatomical axis), and COP excursion percentage, within each exertion direction. Dependent measures were not compared between different exertion directions. A Greenhouse-Geisser correction was used to protect against violations of the sphericity assumption. Pairwise comparisons with a Bonferonni adjustment were used to determine individual differences between different levels of the condition variable. Partial η^2 was used as an estimate of effect size due to the use of a repeated measures design. Alpha was set at 0.05 for all comparisons. All statistical processing was completed using SPSS software (SPSS INC., Chicago, IL, USA).

3.4 – Results

During pulling hand force was significantly affected by the experimental conditions ($p < 0.001$) (Figure 3.3, Table 3.2). When balance was removed as a limiting factor during the braced conditions (i.e. LBB and UBB conditions) hand force capability increased approximately 2.5 times, compared to the SWFP condition. During unilateral medial exertions hand force was also significantly affected by the experimental conditions ($p < 0.001$) (Figure 3.3, Table 3.2). Unilateral medial hand force capability increased 60% when participants were braced as compared to the SWFP condition. Unilateral exertions in a downward direction were also affected by the experimental conditions ($p < 0.001$); however only the force in the UBB trial differed, significantly less than the downward force in the other conditions.

A summary of the ANOVA results (Table 3.2 and 3.3) demonstrates where the experimental conditions had a significant effect on the dependent measures. The significant variables with effect sizes (partial η^2) greater than 0.5 are considered most meaningful (Cohen, 1988) and will be the primary focus of the results.

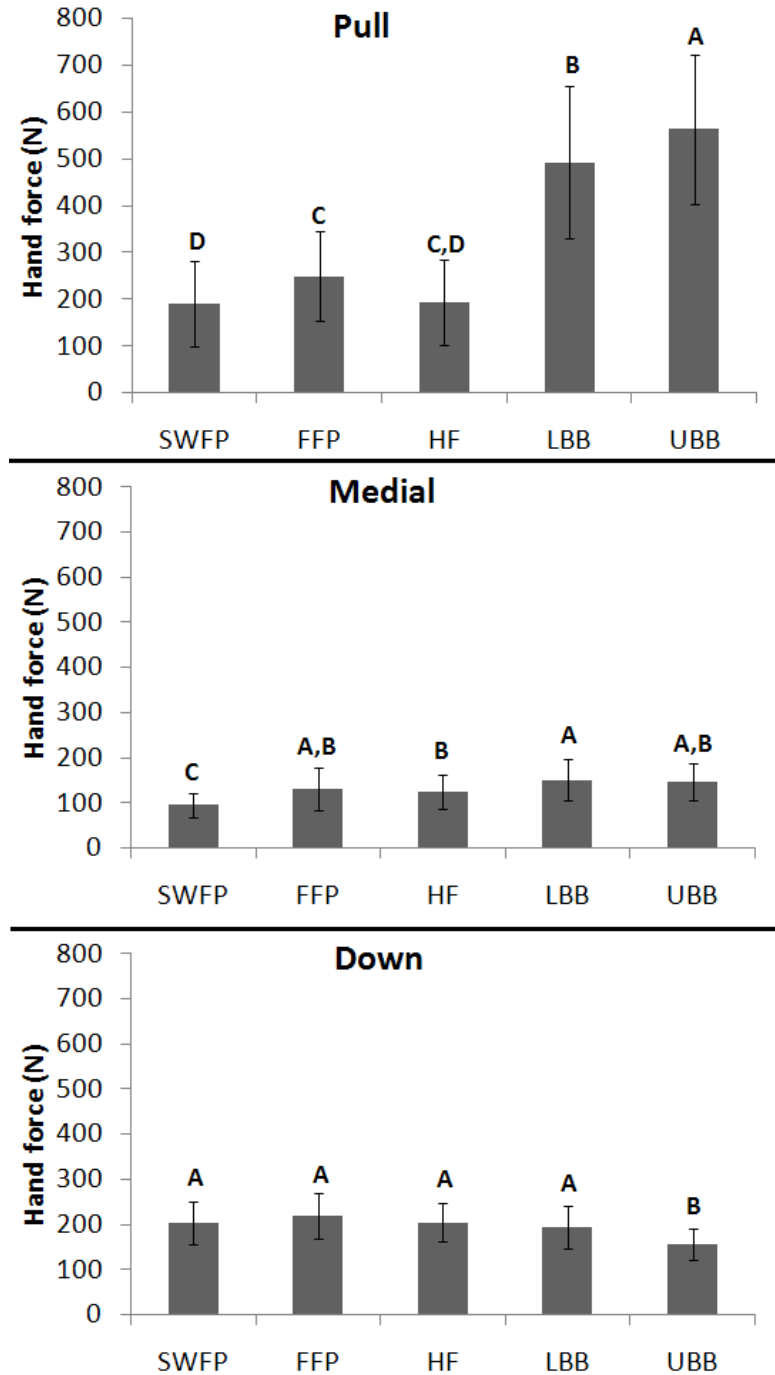


Figure 3.3 – The maximum hand force capability and standard deviation for each experimental condition in each direction. Bars with different letters are significantly different ($p < 0.05$). SWFP = shoulder width foot placement; FFP = free foot placements; HF = high friction; LBB = lower body braced; UBB = upper body braced.

Table 3.2 – Summary of the ANOVA results and the effect of the experimental conditions on the dependent measures: hand force and %MVE for all muscles. Significant findings ($p < 0.05$) and meaningful effect sizes (partial $\eta^2 \geq 0.5$) are bolded and noted with an asterisk (*).

Measure	Pull					Down					Medial				
	df	F	p	Partial η^2	Observed Power	df	F	p	Partial η^2	Observed Power	df	F	p	Partial η^2	Observed Power
Force	1.573	72.241	< 0.001*	0.8*	1.00	2.249	20.856	< 0.001*	0.7*	1.00	2.185	14.295	< 0.001*	0.6*	0.998
Left Rectus Abdominus	1.824	1.133	0.334	0.1	0.217	3.063	0.205	0.896	0.0	0.085	1.687	2.379	0.127	0.2	0.389
Right Rectus Abdominus	1.194	0.989	0.351	0.1	0.163	2.071	0.369	0.703	0.0	0.103	1.361	1.720	0.213	0.1	0.261
Left Erector Spinae	2.462	3.992	0.022*	0.2	0.732	1.554	0.498	0.57	0.0	0.112	2.96	2.44	0.083	0.2	0.551
Right Erector Spinae	2.622	18.485	< 0.001*	0.6*	1.00	2.199	3.093	0.059	0.2	0.566	1.622	1.344	0.281	0.1	0.232
Teres Major	1.773	6.111	0.009*	0.3	0.812	2.09	2.8	0.08	0.2	0.506	1.716	1.188	0.32	0.1	0.216
Latissimus Dorsi	2.091	10.224	< 0.001*	0.4	0.979	2.379	2.036	0.144	0.2	0.414	1.323	0.894	0.389	0.1	0.155
Pectoralis Major - Sternal origin	1.593	1.522	0.241	0.1	0.261	2.741	1.586	0.216	0.1	0.358	1.305	1.198	0.308	0.1	0.191
Pectoralis Major - Clavicular origin	2.331	1.856	0.168	0.1	0.383	1.64	0.924	0.397	0.1	0.173	2.609	1.489	0.241	0.1	0.329
Upper Trapezius	1.923	9.503	< 0.001*	0.4	0.96	1.73	1.996	0.167	0.2	0.338	2.061	0.905	0.421	0.1	0.189
Middle Trapezius	1.801	13.933	< 0.001*	0.5*	0.993	2.403	6.193	0.004*	0.4	0.894	2.247	4.593	0.018*	0.3	0.836

Table 3.3 – Summary of the ANOVA results and effect of the experimental conditions on the dependent measures: COP (M/L and A/P) and the shoulder moments. Significant findings ($p < 0.05$) and meaningful effect sizes (partial $\eta^2 \geq 0.5$) are bolded and noted with an asterisk (*).

Pull					
Measure	df	F	α	Partial η^2	Observed Power
COP - M/L	2.373	10.412	<0.001*	0.5*	0.988
COP - A/P	1.181	140.342	<0.001*	0.9*	1.000
Shoulder Moment (Int/Ext)	2.522	18.336	<0.001*	0.6*	1.000
Shoulder Moment (Add/Abb)	2.759	6.280	0.002*	0.3	0.934
Shoulder Moment (Flex/Ext)	2.373	9.053	<0.001*	0.4	0.977
Down					
Measure	df	F	p	Partial η^2	Observed Power
COP - M/L	2.241	2.063	0.144	0.2	0.404
COP - A/P	1.897	3.561	0.049*	0.2	0.582
Shoulder Moment (Int/Ext)	2.973	13.611	<0.001*	0.5*	1.000
Shoulder Moment (Add/Abb)	2.571	0.421	0.709	0.0	0.120
Shoulder Moment (Flex/Ext)	2.179	9.630	<0.001*	0.4	0.977
Medial					
Measure	df	F	p	Partial η^2	Observed Power
COP - M/L	1.548	10.505	0.002*	0.5*	0.942
COP - A/P	2.237	1.568	0.228	0.1	0.315
Shoulder Moment (Int/Ext)	2.678	9.337	<0.001*	0.4	0.989
Shoulder Moment (Add/Abb)	2.294	17.083	<0.001*	0.6*	1.000
Shoulder Moment (Flex/Ext)	2.734	4.616	0.010*	0.3	0.830

Exposure to different experimental conditions during pulling had a significant and meaningful affect on the %MVE for the middle trapezius ($p < 0.001$, effect size 0.5) and right erector spinae muscles ($p < 0.001$, effect size 0.6) (Figure 3.4). Low effect sizes (expressed as partial η^2) across the remaining %MVE measures suggest that the magnitude of the relationships between the %MVE and exposure to experimental conditions was weak to moderate (Cohen, 1988). Further, the group averaged peak %MVE never exceeded 73%, indicating that no muscle was ever used maximally. Peak %MVE during pulling was recorded from the latissimus dorsi (73%) in the UBB condition; during downward pressing from the teres major (69%) in the FFP condition; and from the pectoralis major – clavicular insertion (70%) during medial exertions in the HF condition. A representative plot illustrating the %MVE for the muscle measured around the torso during downward pressing is provided as trunk muscle strength could be implicated as a limiting factor during downward pressing (Figure 3.5).

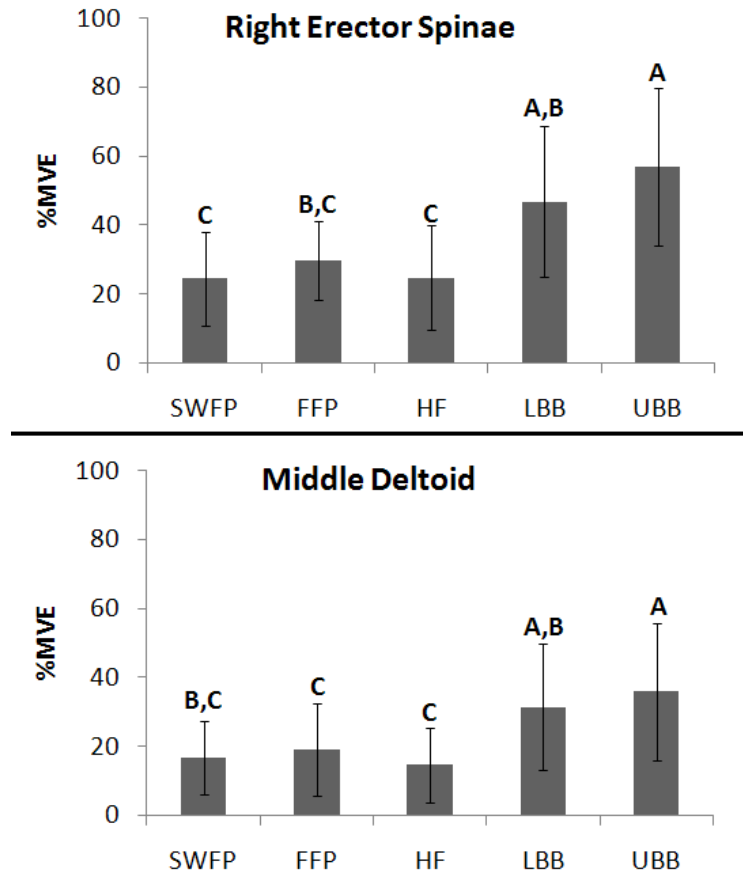


Figure 3.4 – The %MVE and standard deviations for the right erector spinae and middle trapezius during pulling. Bars with different letters are significantly different ($p < 0.05$). SWFP = shoulder width foot placement; FFP = free foot placements; HF = high friction; LBB = lower body braced; UBB = upper body braced.

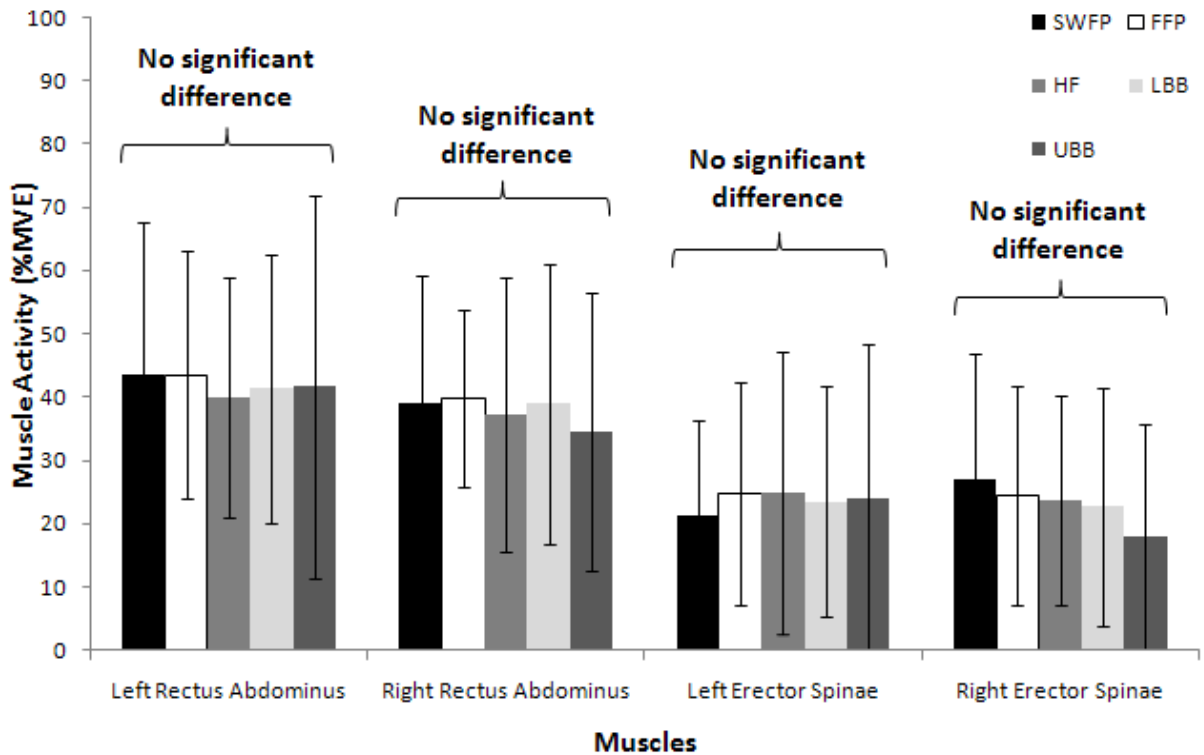


Figure 3.5 – The %MVE recorded from the trunk muscles during unilateral downward exertions. Bars with different letters are significantly different ($p < 0.05$). SWFP = shoulder width foot placement; FFP = free foot placements; HF = high friction; LBB = lower body braced; UBB = upper body braced.

The COP excursion in the A/P direction was both significantly and meaningfully affected during pulling. When pulling, the COP extends forward beyond the BOS during braced conditions (Figure 3.6). The COP excursion in the M/L direction was significantly affected during medial and pulling exertions. Participants chose to shift their COP furthest towards the right side of the BOS when they were braced and less so in the FFP and SWF condition; though they never exceeded the lateral border of the BOS during any exertions (Figure 3.6).

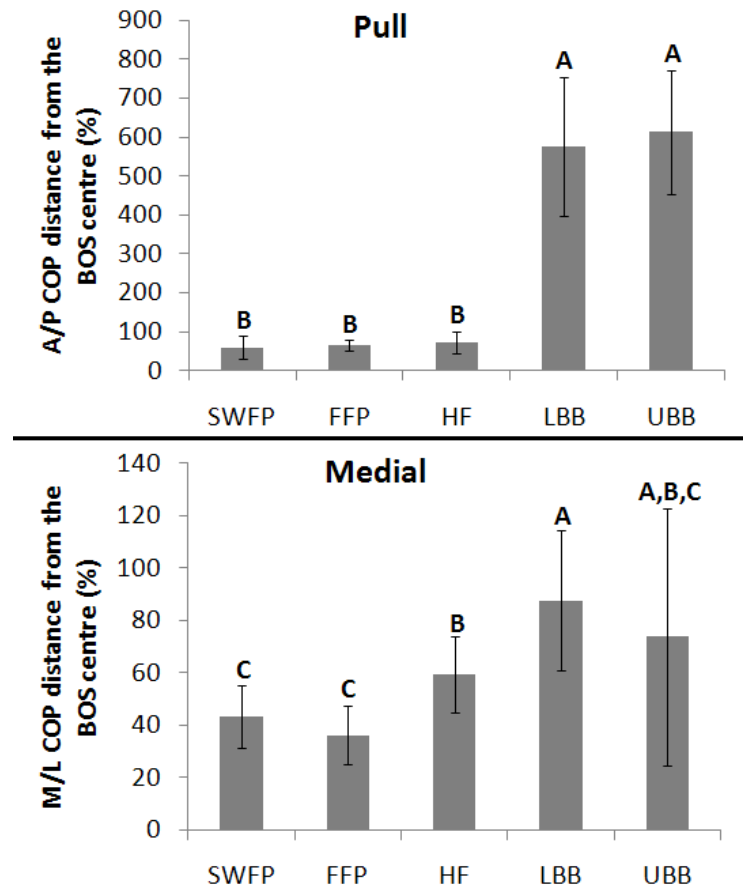


Figure 3.6 – COP distance and standard deviation as a percentage of the distance between the BOS centre and the edge of the BOS. Bars with different letters are significantly different ($p < 0.05$). SWFP = shoulder width foot placement; FFP = free foot placements; HF = high friction; LBB = lower body braced; UBB = upper body braced.

Shoulder moments varied considerably between conditions (Figure 3.7). Shoulder moments in all axes tended to increase as participants were braced during pulling and medially oriented exertions. During downward exertions the adduction moment remained consistent across trials, while the internal rotation moment was decreased significantly when the upper body was braced. The overall peak internal rotation moment occurred during downward pressing in the FFP conditions, while the peak adduction moment occurred during medial pushing in the UBB condition. The peak flexion moment occurred while pushing medially

during the FFP condition, while the peak extension moment occurred during downward pressing in the HF condition.

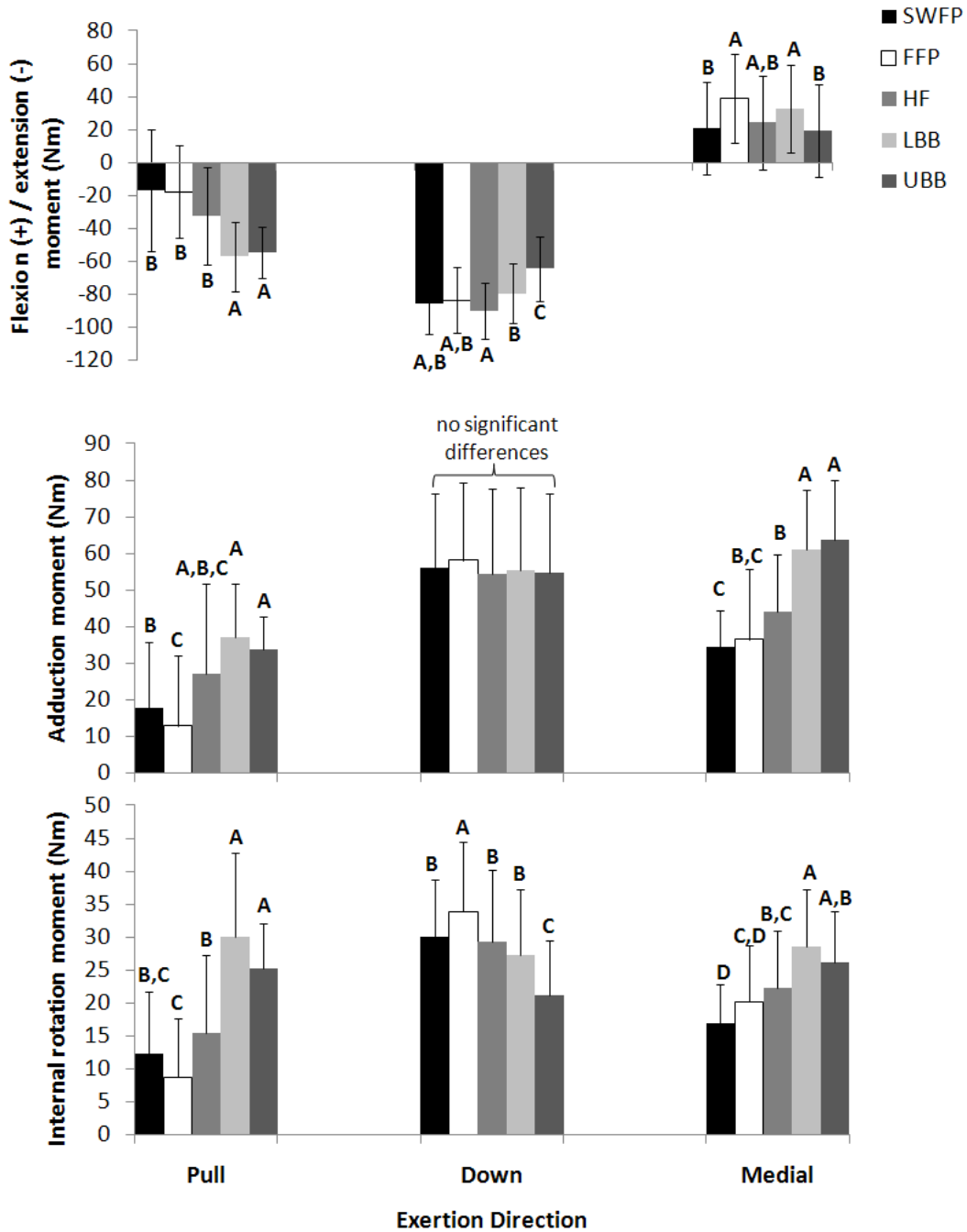


Figure 3.7 – Shoulder moments between conditions. Note that pairwise comparisons were only made between conditions within each exertion direction. Moments were not compared between exertion directions. Bars with different letters are significantly different (p-value < 0.05). SWFP = shoulder width foot placement; FFP = free foot placements; HF = high friction; LBB = lower body braced; UBB = upper body braced.

3.5 – Discussion

The purpose of this study was to identify when specific factors (whole-body balance, shoe-floor friction, and joint strength) limited the ability of the distal upper limb to exert forces on the environment during unilateral pulling, pressing down, and medial exertions. Whole-body balance limited capability during un-braced pulling and medial pushing exertions, while pressing down, braced pulling and medial exertions were limited by joint strength. Shoe-floor friction played a role in medially directed exertions, but was not definitively the limiting factor. However, EMG does not support that these joint strengths are limited by the capacity of individual prime movers acting at the trunk or shoulder given that muscle activations were substantially less than that in the calibration trials. The following discussion will underscore how these conclusions were made based on the results presented, and in context with previous literature.

3.5.1 – *The role of balance as a weakest link*

Balance played the greatest role in limiting exertions in the horizontal plane (pushing medially and pulling inward). The highest unilateral pull and medial push forces were achieved when participants were braced, eliminating balance as a possible limiting factor. This phenomenon can be additionally explained using the COP data. When the participants were not braced, the COP was restricted to the area defined by the BOS to maintain static stability; however, when participants were braced, they allowed their COP to migrate beyond the BOS limits. Further, the COP often remained within a sub-range within the BOS area. This supports the principle of a FBOS region (Holbein and Chaffin, 1997; Holbein and Redfern, 1997; Lee and Lee, 2003; Holbein-Jenny et al., 2007). During pulling in the SWFP condition, the forward excursion of the COP was 59 (± 30) % of the distance from the BOS centre to the

most anterior point. Using a loaded leaning experiment, Holbien-Jenny et al., (2007) reported a comparable average of 58 (± 17) %, where Lee and Lee (2003), using a similar loaded leaning protocol reported a slightly lower value of 51 (± 9) %. The similarity between the FBOS reported previously and the COP excursion calculated during pulling exertions in this study are consistent with balance being the most likely limiting factor. Whether leaning as far as possible or pulling as hard as possible, a similar A/P limit was reached.

COP excursions in the M/L direction during medial exertions were not as consistent as those in the A/P direction during pulling. Hobein-Jenny et al., (2007) reported an M/L FBOS of 66 (± 12) % while Lee and Lee (2003) reported an M/L FBOS of 58 (± 12), both more than the M/L COP excursion in this study: 43 (± 12) %. This may indicate that M/L balance was not limiting in the current study. As indicated in Figure 3.3, removing balance as a constraint through bracing did result in an increase in force relative to the SWFP condition; however the M/L COP excursion was only significantly greater during the LBB braced condition and not the UBB braced condition. These experimental results suggest that M/L balance may play a role in the production of unilateral medial pushing forces, but that role is unclear in the present study. Though it is puzzling that participants did not reach the limits described in the literature, it may be that the M/L forces cause a rotational moment about the vertical axis projected up from the COP. This rotational moment may be internally controlled similarly to A/P or M/L balance, but was not monitored in the current study. It may also be possible that our participant pool had different ankle strengths, or postural control strategies (Holbein and Redfern, 1997) than those observed previously, reducing the effective FBOS range in the M/L direction for our participant pool.

3.5.2 – The role of joint strength as a weakest link

Joint strength was most likely to limit unilateral hand force capacity in the absence of a balance or friction related constraint. The downward pressing forces measured in this study were not limited by balance as force capability was consistent across exertions, and did not increase when the possibility of a balance limitation was removed by bracing. Since force capability could not be increased by eliminating balance and friction based constraints, joint strength emerges as the limiting factor in this direction (Figure 3.6). Shoulder and trunk joint strength were considered, a priori, to be most likely to limit downward pressing in the absence of a balance or friction constraint, based on the moment arms between the point of force application and each of those joint centres. With respect the shoulder joint the peak internal rotation (34 ± 10 Nm) and extension moments (90 ± 17 Nm) calculated during downward pressing approached estimated population maximum internal rotation (37 ± 9 Nm) and extension (97 ± 20 Nm) strengths predicted using the equations presented by Schanne (1972) and Stobbe (1982) incorporated into the 3DSSPP. These data suggest that shoulder strength may limit unilateral hand force during downward exertions.

Trunk moments were not calculated in this study due to the use of the bracing apparatus. Instead EMG measured bilaterally from the rectus abdominus and the erector spinae were used to examine whether trunk strength could be limiting. As indicated in the Figure 3.5, none of the torso muscles recorded reached more than 50% activation during downward exertions relative to the activity recorded during the standardized MVIE tests. However, many other muscles act to move and control the trunk, and may have reached near maximum activity, but have gone unnoticed due to this study design. Kerk et al., (1998) calculated trunk moments during downward pressing in a similar posture and then compared those moments to the population strength data described above. They reported that trunk

moments often reached population thresholds, rather than moments at other joints, supporting a trunk strength limitation during downward exertions. They further hypothesized that strong subjects should maintain an advantage over weaker subjects when exerting force in the vertical direction due to the underlying strength limitation, while stronger and weaker subjects should respond similarly during horizontal pulling where balance is typically more limiting (Kerk et al., 1998).

By default, joint strength was the limiting factor during all UBB exertions. Similar to the increases in force that occurred with bracing, shoulder joint moments increased considerably from non-braced conditions, approaching but not reaching the same magnitude observed during downward exertions and presented by previous authors. Medial exertions also resulted in a maximum adduction moment of 64 ± 16 Nm, lower than the maximum adduction moment 88 ± 24 Nm predicted using the equations presented by Schanne (1972) and Stobbe (1982). This suggests that another joint may have reached capacity before the shoulder moment capacity was reached. For instance, trunk strength may have limited these exertions. Previous research suggests that participants attempt to minimize trunk moments during pushing and pulling tasks (de Looze et al., 2000; Hoozemans et al., 2004; Hoffman et al., 2007), which is consistent with the trunk being a weak link. In the current study, trunk moments were not determined during the braced conditions and force was not recorded between the participant and the bracing apparatus.

3.5.3 – Is joint strength limited by the individual capacity of a prime mover?

Electromyography provided quantification of the muscle activation of primary movers during the various exertions. If joint strength is limiting, it could be expected that muscle activity would also increase to near 100% MVE in one or all of the primary movers associated

with that joint. However, few muscles were significantly or meaningfully affected by changing conditions (Figure 3.4, 3.5); which suggests that muscle activity in the primary movers was not highly influenced whether a joint was limiting or not. This implies that either a secondary muscle not recorded in this study has achieved a maximum, therefore limiting joint strength, a different joint was limiting where muscles were not monitored (ankle, hip or knee) or a more complex phenomenon exists to set the upper limit for joint strength. For instance, joint stability maintenance may be intrinsically responsible for maximum joint strength when a joint limits hand force. During maximum pushing and pulling tasks, Granata and Bennett (2005) noted that participants attempted to limit trunk moments. In doing so, the corresponding lower moments resulted in lower trunk stability and therefore much higher levels of trunk muscle co-contraction were required to compensate for the decreased stability (Granata and Bennett, 2005). In the study by Granata and Bennett (2005) joint strength was not limited by the ability the prime movers to produce an opposing moment, but rather they were limited in their ability to remain stable while producing the opposing moment. At the shoulder Veeger and van der Helm (2007) note that the large primary movers can help provide stability, but they can often result in large antagonistic moments. Therefore, the result of a 69% active teres major during a shoulder limited downward exertion, for example, could require near 100% activity from corresponding rotator cuff muscles to ensure the joint remains in a stable position within the glenoid cavity. Therefore, although this research has determined that joint strength can limit the ability of the distal upper limb to exert forces the data does not support that this joint strength limitation is related to the muscle activation capacity of the prime movers.

3.5.4 – Limitations

This study had inherent limitations. Only one handle location relative to the participants was tested. Errors arising from the postural data collection, such as inter-trial skin motion and marker placement accuracy, and the use of population-based anthropometric tables to estimate segment masses, cause some uncertainty in the COP calculation and the linked segment modeling outcomes. These effects were mitigated through the use of rigid clusters to help reduce potential artifacts in the motion capture data (Kingma et al., 1996) and the error from the use of anthropometric tables would likely be randomly distributed across participants. Only male university students participated in this study. Males were selected for ease of electrode placement and bracing options, whereas female participants would have required the upper body brace to be placed higher on the chest (rather than at the nipple line) and the area of the suprasternal notch could not be covered as reflective markers were positioned there and were to remain visible.

3.6 – Conclusions

This study examined potential hand force limiting factors or constraints during inward pulling, medial pushing, and downward pressing and attempted to explain why these factors were potentially limiting. Unilateral pulling hand forces were limited by whole-body balance. Downward pressing was not limited by balance and may be more likely limited by another factor, potentially trunk or shoulder joint strength. Medial exertions were affected by balance, and shoe-floor friction; however, it was not clear which factor is the greatest limiter. When joint strength was identified as the likely constraint no individual primary movers of the shoulder or trunk reached maximal activity, suggesting that the joint strength limitation may be guided at a deeper level by internal joint requirements, such as joint stability, or the joint strength limitation may occur at joints where corresponding muscles were not monitored (i.e.

lower body). This study concludes that foot placement, handle height, distance from the handle, friction, and body posture can all influence force producing capability, as each of these factor affects the required balance needs and joint moment capacities that may constraint performance.

Chapter 4 – A stochastic 3D static hand force prediction model for estimating maximum feasible hand forces during unilateral exertions: Part 1 – Motivation and model description

Steven L Fischer, Clark R Dickerson and Richard P Wells

4.1 – Overview

Accurate estimation of occupational performance capacity facilitates better proactive job design, or reactive redesign, by informing appropriate changes to job demands. This chapter (Part 1 – Motivation and model description) explains the rationale and describes a three-dimensional biomechanical hand force prediction model to estimate the maximum hand force capability during unilateral upper limb exertions. The model includes a novel probabilistic approach for predicting hand force capacity. The inclusion of a stochastic method provided an opportunity to evaluate maximum force demands in a population context, using a percent capable approach and identifies both a primary limiting factor, and a description of the probability that other factors may be limiting performance.

The main function of the model is to identify the biomechanical factor most likely limiting hand force capacity, then to determine the maximum hand force capacity possible given that limiting constraint. Probability was included by randomly selecting constraint thresholds from a normal distribution of probable thresholds and predicting the corresponding hand force that would result. Monte Carlo simulation was used realize the likely distribution in maximum hand force.

The following chapter (Part 2 – A model evaluation) provides an evaluation of the model. The evaluation was conducted by comparing model predictions with experimental results from a laboratory strength study where participants exerted hand forces chosen to be limited by balance, floor friction and joint strength.

4.2 – Introduction

Overexertion injuries can diminish both worker productivity and quality of life. In Ontario in 2008, overexertion injuries represented 28.1% of all lost time claims paid out by the Workplace Safety and Insurance Board, the highest percentage of any injury event (WSIB, 2008). Further, from 1999-2008 this percentage has fluctuated between 29.1 and 27.4% (WSIB, 2008). Conceptually, these injuries result from excessive exertion which exceeds the tolerance limit or capacity of the system or system components (McGill 1997; Kumar, 2001). Therefore, reducing the incidence of these injuries requires robust limits or capacity thresholds, to help identify if an exertion poses unacceptable risk. A better understanding of hand force exertion capacity may help guide more effective work designs to decrease the occurrence of excessive exertions.

Three approaches have been developed to address force exertion capacity during occupational exertions. The Michigan three-dimensional static strength prediction program (3DSSPP) (summarized in Chaffin, 1997) is the most widely adopted in the field (Dempsey et al., 2005). The program does not predict force exertion capacity directly; but rather it computes the percentage of the population that has the necessary joint strength capacity to perform a given exertion. The model requires the user to input a force and application direction, in addition to posture and anthropometric data to predict the corresponding population strength capacity. However, it may be more favorable to predict the force as an output. This would reduce the number of inputs required to drive the model, and improve its usefulness in a proactive scenario where force requirements may not be known.

A second approach uses the postural stability diagram (PSD) which provides a graphical approach for predicting force exertion capacity (Grieve, 1979a; 1979b). This method

is used to identify which mechanical constraints may be operational for a specified exertion (Grieve, 1979a), revealing the maximum force capacity. The approach is founded upon the equation for static equilibrium for a given exertion, based on the forces acting at the hands and at the feet. The equation is then used to define the relationship between the horizontal and vertical force components, which are then plotted on the PSD in the horizontal and vertical axis respectively. For example, assuming a task requires a horizontal push ($F_{horizontal}$), the following equation for linear static equilibrium would apply:

$$F_{horizontal_hand} = (F_{vertical_hand} + F_{vertical_feet}) * \mu + F_{horizontal_feet} \quad (\text{eq. 4.1})$$

Where F is the force acting in the vertical and horizontal directions at both the hands and feet, and μ is the coefficient of friction. The net vertical and horizontal forces can then be plotted as bound by their relationship to the μ , defining the range of possible forces. This requirement to determine the relationships each time, depending on the work environment, limits the usefulness of the PSD as an applied tool. Further the analysis capacity is restricted to two-dimensional symmetrical tasks and it does not consider strength as a possible limiting factor.

Thirdly, a two dimensional static human force exertion capability model (2DHFEC) was developed drawing on concepts from the previous two models (Kerk et al., 1994). The 2DHFEC predicts maximum hand force capability in the prescribed direction based on the input of anthropometric and posture data. Similar to the 3DSSPP, the 2DHFEC is based on a biomechanical inverse static model; which incorporates population joint strength profiles to constrain individual joint moments during maximum hand force predictions. Similar to the PSD, it also incorporates shoe-floor friction and whole body balance as constraints to hand force predictions. The model is advantageous in that force is an output rather than an input, thus reducing the number of inputs to run the model. However, the utility of the model is

limited to 2D sagittal plane exertions and moreover the model does not provide information about population variability.

A force prediction model could be improved by combining three dimensional analysis capability and population scalability (similar to 3DSSPP), with the prediction of force as an output variable (similar to the PSD and 2DHFEC). Three-dimensional analysis capability increases the range of tasks for which a model could be used to generate estimates. It requires the input of more detailed posture information to describe the third dimension,

Population variability considerably affects exertion capability (Chaffin, 1997). The variability in exertion capacity may be due to the well established population variability in underlying joint strengths (Stobbe, 1982; Kumar, 1996) and balance control (Holbein and Chaffin, 1997; Holbein and Redfern, 1997; Lee and Lee, 2003; Holbein-Jenny et al., 2007). Including this type of variability in a model could help improve predictions, and it may help explain population variability in hand force capability. A probabilistic modeling approach can be introduced to model this variability. Similar probabilistic approaches have shown success when defining a reachable workspace (Venema and Hannaford, 1991), determining the net present value of cash flow resulting from low back pain interventions (Hughes and Nelson, 2009), and predicting external rotation strength of the shoulder (Langenderfer et al., 2006).

The objective of this two part paper is to describe (Part 1) and evaluate (Part 2) a novel model to predict maximum hand force capability during unilateral pulling, downward pressing and medial pushing tasks. It includes a three-dimensional analysis and a probabilistic approach for modeling the constraints that can limit hand force capacity. Part 1 provides a detailed description of the model. Part 2 presents an evaluation that compares model predictions with maximum forces obtained experimentally for both constrained and unconstrained exertions.

4.3 – Methods

4.3.1 – A general overview of the model

The Three-Dimensional Hand Force Prediction Model (3DHFPM) was developed to predict maximum unilateral hand force capability during occupational tasks (Figure 4.1). The model was derived using a weakest link strategy, where the maximum hand force was determined as the highest force capable without exceeding the weakest constraint. A probabilistic approach was used to generate constraint thresholds based on literature data. The model output was a distribution of maximum hand forces based on the constraints and a description of the most frequently selected constraint, or weakest link. The model hierarchy is provided in Figure 4.2 and each component is described in more depth in the following sections.

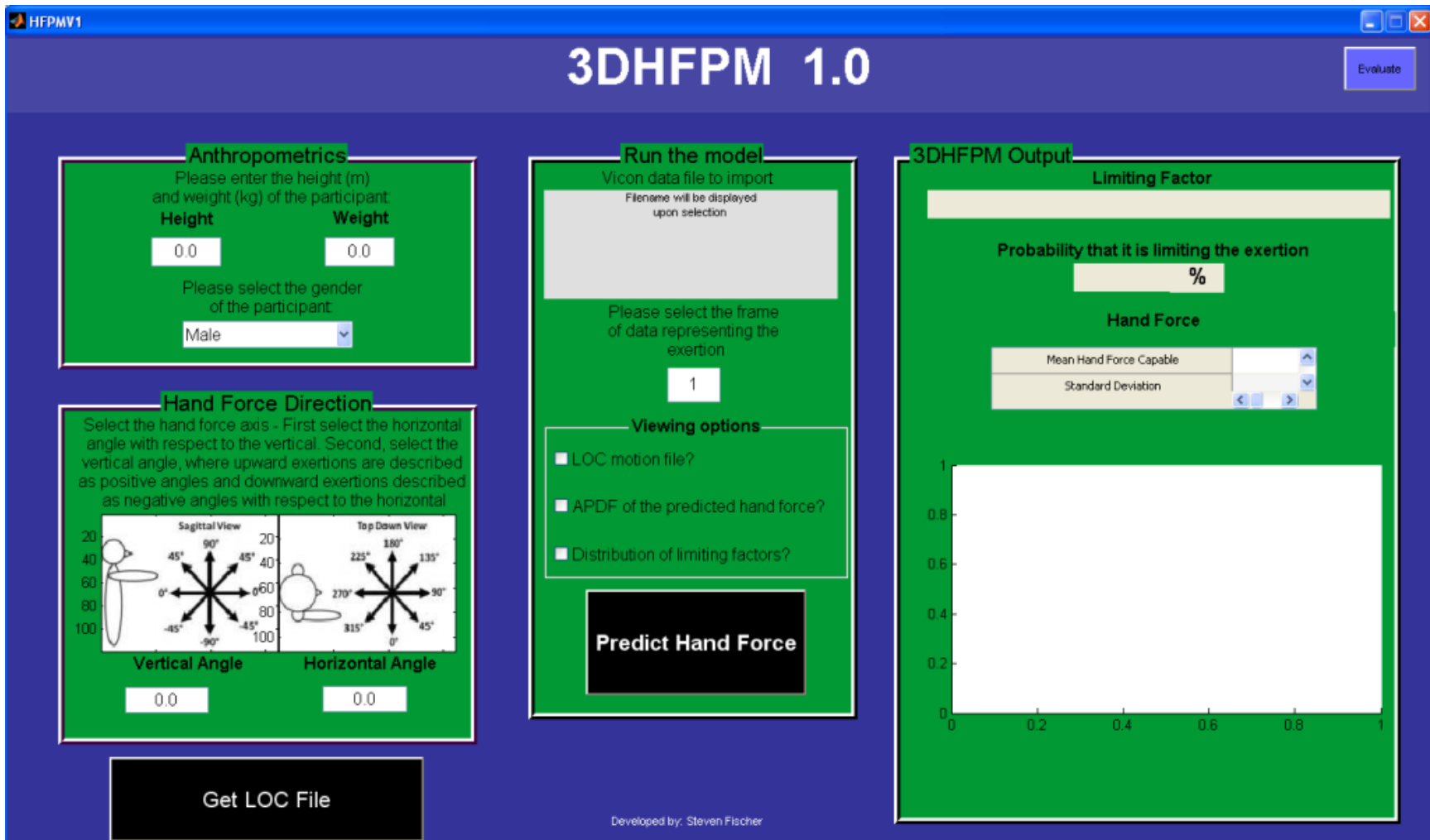


Figure 4.1 – The user interface for the 3DHFPM. Input data includes: participant stature and body mass, information on hand force direction, and a posture (“Get LOC File” – where an LOC file is a lab specific standardized method for describing posture data obtained using motion capture equipment).

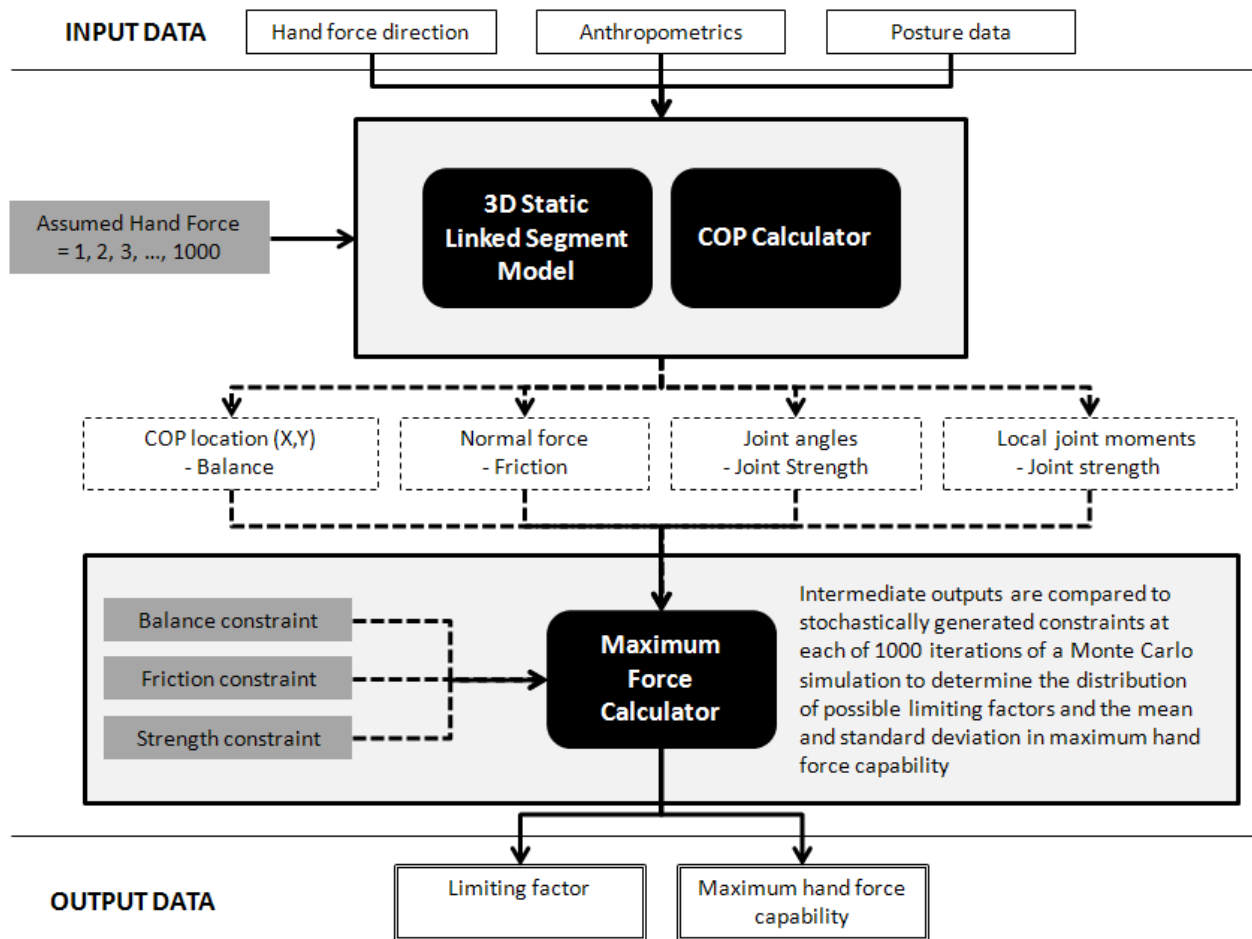


Figure 4.2 – The model hierarchy and data flow through the components of the model. Required inputs (framed with a single solid line) include hand force direction, anthropometrics and posture. Intermediate outputs (framed with dashed lines) include the COP location, the normal force, joint angles and the local joint moments. The final outputs (framed by double lines) include the most likely limiting factor and the maximum hand force capability. Major model components are highlighted in black with white text.

Since hand force is not an input, but is required for estimating the joint moments, COP location and frictional force, the initial hand force was assumed to be equal to 1 N and was iteratively increased by increments of 1 N up to 1000 N. The resulting joint moments, COP locations and frictional forces for each hand force magnitude were compared to the constraint thresholds for each measure to determine at which level of hand force the first constraint was exceeded. The static model is based a single instantaneous posture, which assumes that this

posture represents the posture required to produce a maximum force in the prescribed direction for a person with those anthropometric attributes.

4.3.2 – Three-dimensional static linked segment model

The linked segment portion of the model served to calculate five parameters: the whole body center of mass location, potential hand forces, the normal force, joint angles at the elbow, shoulder and trunk, and joint moments at the elbow, shoulder and trunk (Figure 4.3). The linked segment model was described using seventeen segments (Figure 4.4). The anthropometric segment specific properties included: mass and COM locations and were described using the ratio relationships presented by Winter (2005). The left and right clavicle segments were assumed to be mass-less. A geometric heuristic approach was used to estimate the joint centers from the motion capture data for the glenohumeral (represented by the center of the humeral head), elbow, and the wrist joints (Nussbaum and Zhang, 2000) in addition to the trunk endpoints (c7 and L5 vertebra) (Dickerson, 2005). Nominal joint centre locations were estimated for the ankle and knee (Zatsiorsky, 1998) and for the hip joint centre (Bell, 1989). Using the joint center locations, the whole body center of mass location was calculated using equations presented by Winter (2005).

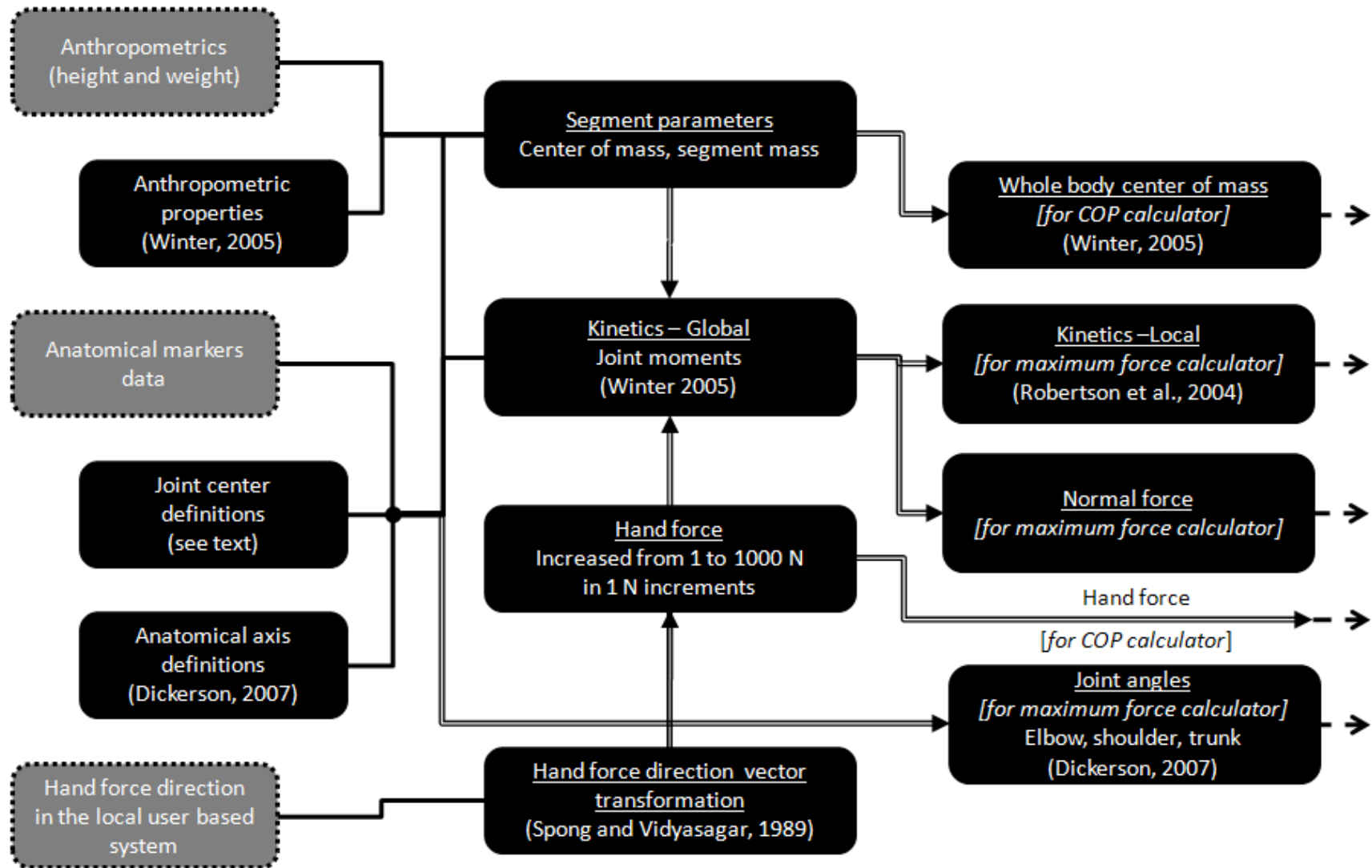


Figure 4.3 – A schematic representation of the input (grey boxes) and calculations in the 3D static linked segment model.

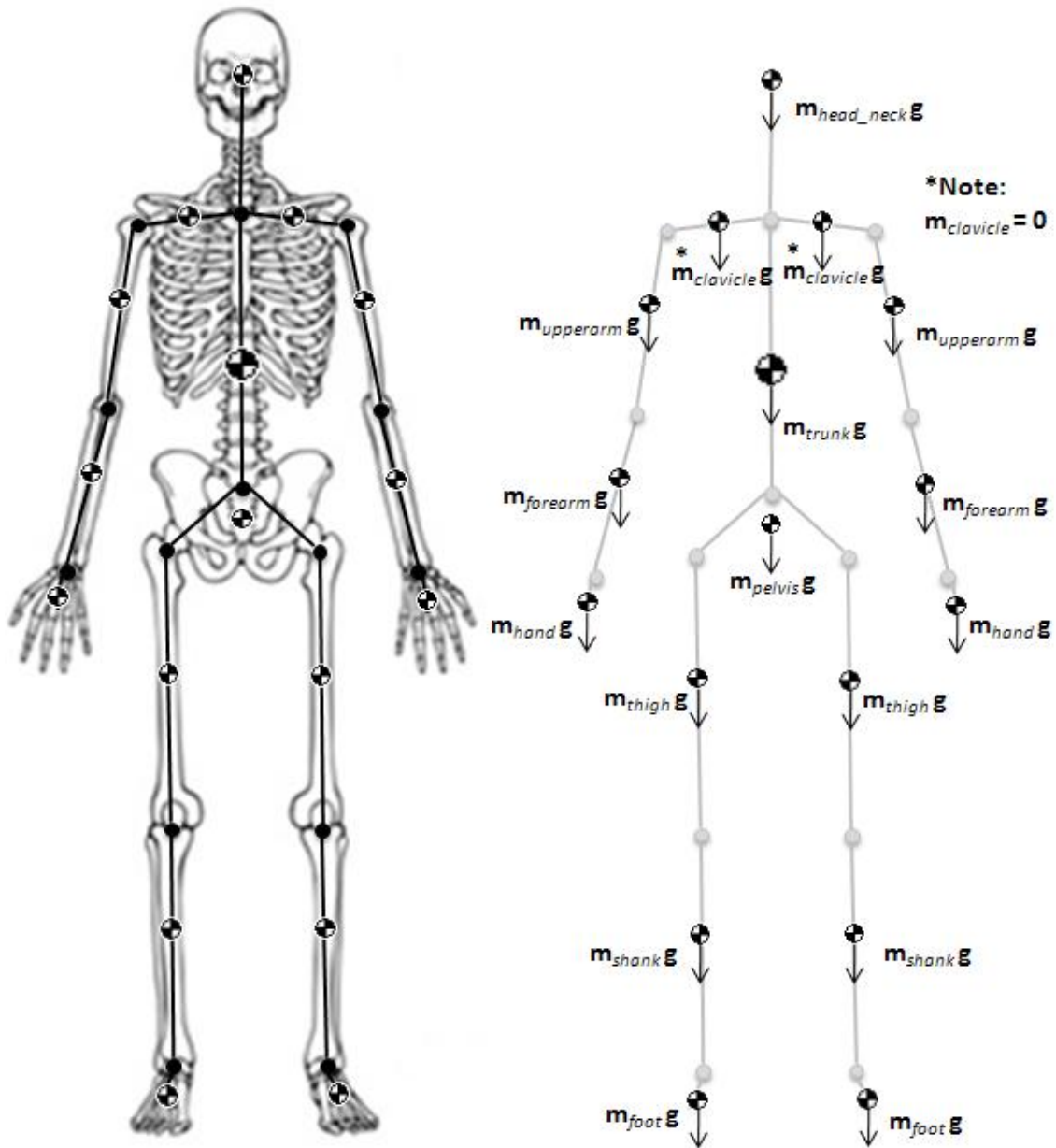


Figure 4.4 – An illustration of the 17-segments used in the 3DHFPM. In the stick figure on the right, $m_{\text{subscript}}$ refers to the segment mass, and g refers to the acceleration due to gravity.

4.3.2.1 – Anatomical axis system definitions

Anatomical axis systems were defined for the forearms, upper arms, trunk and pelvis (Figure 4.5), adapted from Dickerson et al. (2007).

The forearm system had its origin at the elbow joint center. The positive x-axis was directed through the wrist joint center. The positive z-axis was described as the cross product of the forearm x-axis and the upper arm x-axis, directed laterally. The positive y-axis was orthogonal to the x- and z-forearm axes.

The upper arm system originated at the center of the humeral head. The positive x-axis was directed through the elbow joint center. The positive z-axis was described as the cross product of the forearm x-axis and the upper arm x-axis, directed laterally. The positive y-axis was orthogonal to the x- and z-upper arm axes.

The trunk system originated at the L4/L5 vertebra. The positive x-axis was directed through the C7 vertebra. The positive z-axis was described as the cross product of the trunk x-axis and a vector extending from the C7 vertebra to the suprasternal notch marker. The z-axis was directed to the left side of the body. The y-axis was orthogonal to the x- and z-trunk axes.

The pelvis system originated at the midpoint between the left and right hip joint centers. The positive x-axis was directed through the L4/L5 vertebra. The positive z-axis was described as the cross product of the pelvis x-axis and a vector extending from the L4/L5 vertebra to the midpoint between the anterior superior iliac spine markers. The z-axis was directed to the left side of the body. The y-axis was orthogonal to the x- and z-axes.

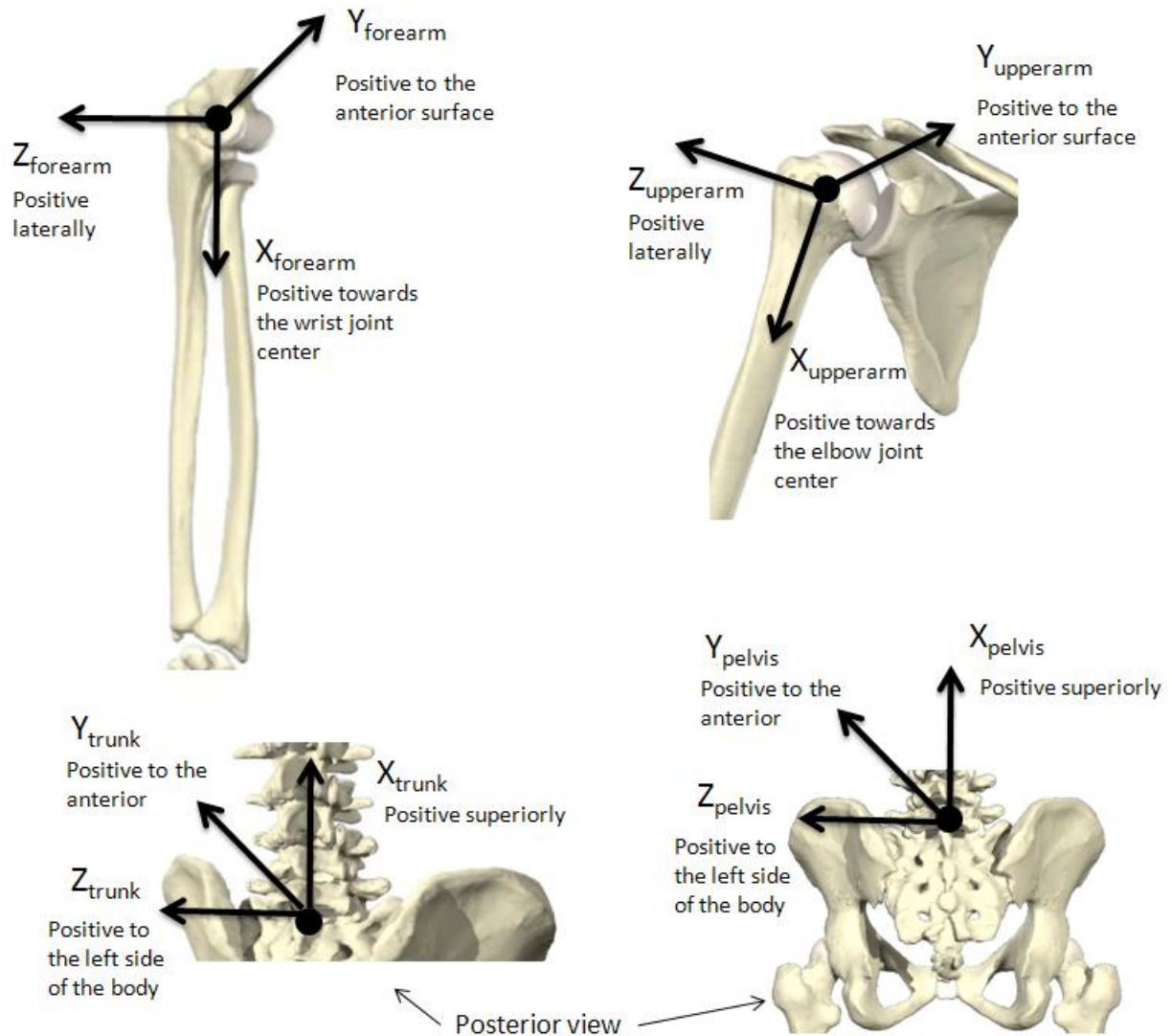


Figure 4.5 – Anatomical axis definitions for the forearm, upper arm, trunk and pelvis. Skeleton graphics were created using Primal Pictures ©2010.

4.3.2.2 – Inverse static calculations

A top down inverse static analysis procedure was used to calculate joint moments in global coordinates at the elbow, shoulder and trunk. Each body segment is considered a free body diagram (Figure 4.6), interconnected at the joint centers (Kingma et al., 1996).

Newtonian mechanics were used to solve for the moments using equations of static equilibrium. Normal force acting at the foot-floor interface was also calculated, as the

summation of the force due to gravity acting on the body and any vertically directed hand force components.

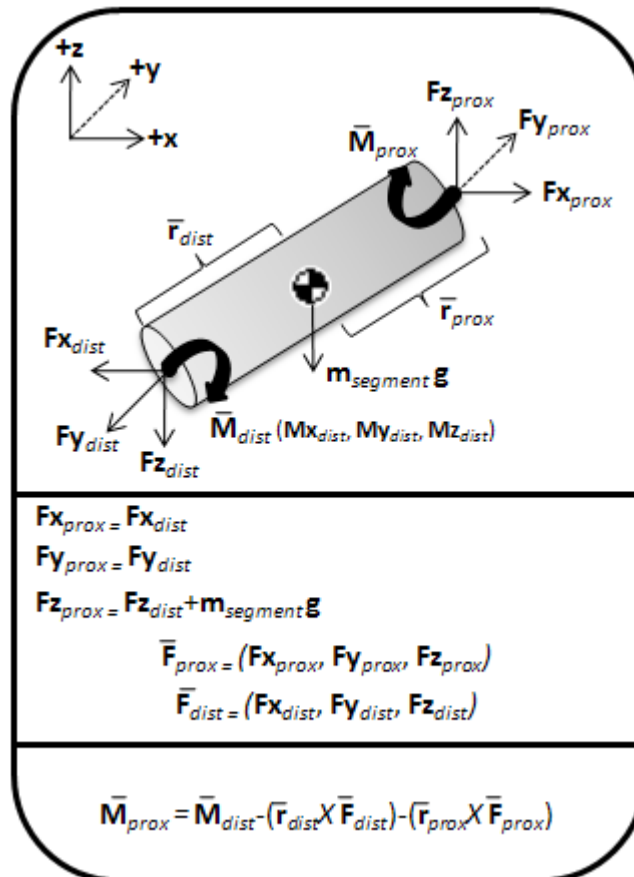


Figure 4.6 – A representative free body diagram and subsequent equations of static equilibrium used to calculate joint moments at the elbow, shoulder and trunk.

To drive the top-down inverse static approach the force magnitude and application direction were required. Since force magnitude was a desired output, it was assumed to be equal to 1, then 2, and so on up to 1000 N, with the corresponding resultant data being stored for later use in the maximum force calculator component. The force application direction is a required input. The application direction is entered with respect to two pre-defined axis – a horizontal angle, measured about the global z axis (in line with gravity), and a vertical angle, measured about the global x-axis (configured to be in the medial/lateral direction for this

study). A three-dimensional force direction unit vector was then determined by rotating a vector [1, 0, 0] through the horizontal then vertical angles using a method outlined by Spong and Vidyasagar (1989). Subsequent possible net hand forces were calculated assuming hand force was 1 to 1000 N by multiplying the possible force by the directional unit vector.

4.3.2.3 – The consideration of off-axis forces

Several researchers have reported considerable off-axis force production during maximum exertions in addition to the force produced in the prescribed direction (Abel et al., 1991; Fothergill et al., 1991; de Looze, et al., 2000; Granata and Bennett, 2005; Hoozemans et al., 2007; Hoffman et al., 2007). The additional off-axis forces alter the direction of the force vector from the prescribed direction. Current theory suggests that these off-axis forces are produced to help reduce the joint loads at the shoulder and low back (de Looze et al., 2000; Granata and Bennett, 2005; Hoffman et al., 2007). The current model does not account for any potential shifting of the force direction vector, and assumes that force is produced in the user prescribed direction. To determine the effect of this assumption, the model was evaluated in its current state (assuming no change in the force vector direction), and again by revising the prescribed direction to match the actual direction (See Figure 4.7 for a representative example using a pull exertion). The actual force vector was extracted from the experimental force data for each exertion, generated to evaluate the model.

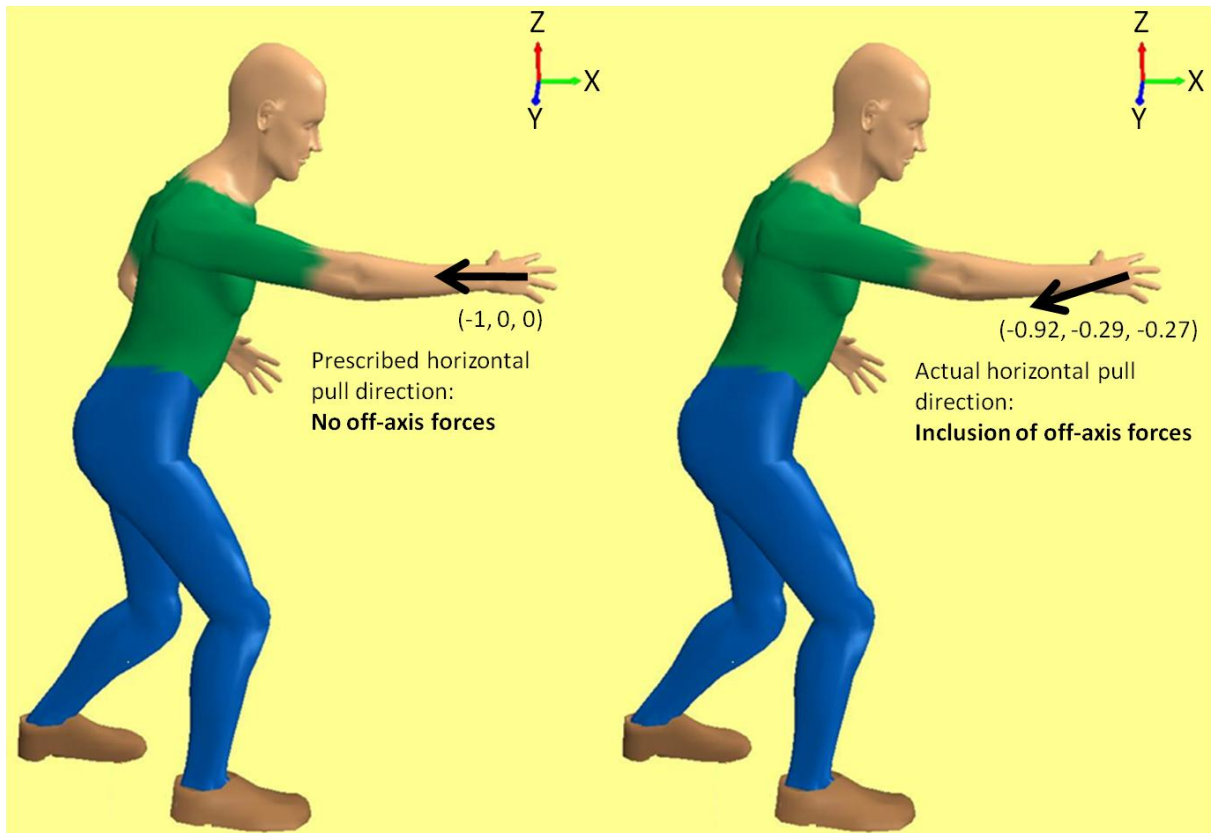


Figure 4.7 – An illustration to demonstrate how off-axis forces impact the actual force direction angle using a pulling example. Based on data from a typical participant, when the off-axis forces are considered, the actual net force direction shifts from the prescribed direction. *Note: the hominoid graphic was generated using the 3DSSPP™.

4.3.2.4 – Transformation matrices and Cardan angle decomposition

In addition to calculating the whole body COM location, the hand force vector, and the normal force, two remaining intermediate outputs were required for subsequent model components: 1) the local joint moments at the elbow, shoulder and trunk and 2) joint angles for the elbow, shoulder and trunk respectively. Local joint moments were calculated in a two step process using an approach presented by Spong and Vidyasagar (1989). First segment specific transformation matrices (T) were derived to describe the relationship between the global coordinate system (GCS) and the segment specific local coordinate systems (LCS):

$$\begin{bmatrix} T_{LCS}^{GCS} \end{bmatrix} = \begin{bmatrix} i_{GCS} \cdot i_{LCS} & j_{GCS} \cdot i_{LCS} & k_{GCS} \cdot i_{LCS} \\ i_{GCS} \cdot j_{LCS} & j_{GCS} \cdot j_{LCS} & k_{GCS} \cdot j_{LCS} \\ i_{GCS} \cdot k_{LCS} & j_{GCS} \cdot k_{LCS} & k_{GCS} \cdot k_{LCS} \end{bmatrix} \quad (\text{eq. 4.2})$$

The local joint moments (M_{LCS}) were then determined using the transformation matrices and the global joint moments (M_{GCS}) by using the following equation:

$$M_{LCS} = \begin{bmatrix} T_{LCS}^{GCS} \end{bmatrix} M_{GCS} \quad (\text{eq. 4.3})$$

The joint angles were similarly calculated during a two step process. First transformation matrices were derived to describe the relationship between proximal and distal segments, LCS_1 and LCS_2 :

$$\begin{bmatrix} T \end{bmatrix} = \begin{bmatrix} i_{LCS_1} \cdot i_{LCS_2} & i_{LCS_1} \cdot j_{LCS_2} & i_{LCS_1} \cdot k_{LCS_2} \\ j_{LCS_1} \cdot i_{LCS_2} & j_{LCS_1} \cdot j_{LCS_2} & j_{LCS_1} \cdot k_{LCS_2} \\ k_{LCS_1} \cdot i_{LCS_2} & k_{LCS_1} \cdot j_{LCS_2} & k_{LCS_1} \cdot k_{LCS_2} \end{bmatrix} \quad (\text{eq. 4.4})$$

Then, using a Cardan z-y-x decomposition, similar to Dickerson et al. (2007) the corresponding angles were determined. The first angle (α) was about the flexion / extension axis (z), the second (β) about the adduction and abduction axis (y) and the third (γ) about the segmental axial twist axis (x):

$$\begin{bmatrix} T \end{bmatrix} = \begin{bmatrix} R_x \end{bmatrix} \begin{bmatrix} R_y \end{bmatrix} \begin{bmatrix} R_z \end{bmatrix} \quad (\text{eq. 4.5})$$

$$\begin{bmatrix} T \end{bmatrix} = \begin{bmatrix} \cos \alpha \cos \beta & \sin \alpha \cos \beta & -\sin \beta \\ (-\sin \alpha \cos \gamma + \cos \alpha \sin \beta \sin \gamma) & (\cos \alpha \cos \gamma + \sin \alpha \sin \beta \sin \gamma) & \cos \beta \sin \gamma \\ (\sin \alpha \sin \gamma + \cos \alpha \sin \beta \cos \gamma) & (-\cos \alpha \sin \gamma + \sin \alpha \sin \beta \cos \gamma) & \cos \gamma \cdot \cos \beta \end{bmatrix} \quad (\text{eq. 4.6})$$

The local joint moments at the elbow, shoulder and trunk, in addition to the normal force, were all calculated and stored as hand force increased in increments of 1 N from 1 to

1000 N. The corresponding net hand forces were also stored for use in subsequent model components.

4.3.3 – COP calculator

The center of pressure (COP) was also calculated using a top down approach. The data used to drive the calculation included: whole body COM location, the hand force and the global location of the point of application of the hand force (assumed to be at the COM of the hand). The COP location was calculated at each increment of hand force. The corresponding COP locations could then be compared in the Maximum Force Calculator to determine which level of hand force caused the COP to migrate beyond the functional base of support (FBOS). The COP location was determined by solving for the static moment equilibrium about the COP location in the global X and Y directions. The A-P COP location is dependent on the components of force applied in the vertical (RF_z) and along the AP horizontal axis (RF_y), while the M-L COP location is again dependent on the force applied in the vertical (RF_z) axis, but also along the ML horizontal axis (RF_x). The vertical location of the COP is assumed to always be on the floor, and no axial rotation moments about the COP were determined. By simplifying the AP and ML COP calculation into two separate 2D analyses, the corresponding moment arms required to solve the AP and ML moment equilibrium equations could be obtained from the marker data. The diagram (Figure 4.8) demonstrates how the COP in the global Y direction was calculated using a pulling example.

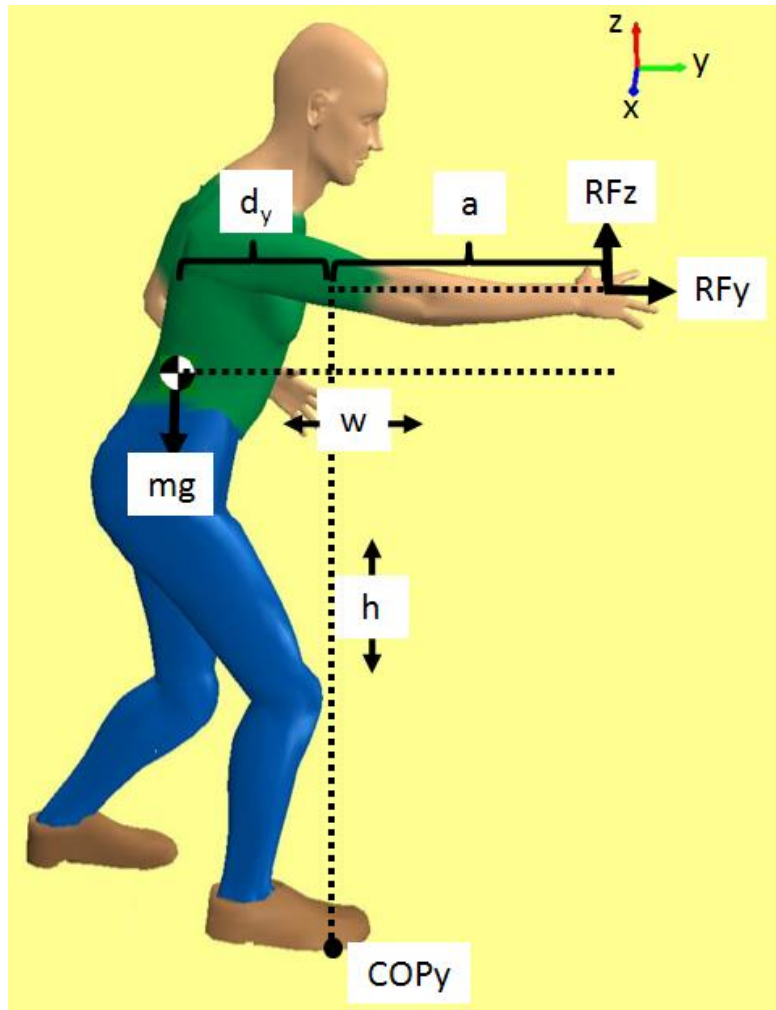


Figure 4.8 –Sagittal plane diagram used to demonstrate how the static moment equilibrium about the COP in the global X axis was determined. The specific variable definitions are described below following the derivation of the equation. *Note: the hominoid graphic was generated using the 3DSSPP™.

$$\begin{aligned}
 0 &= mgd_y + (RF_z * (w - d_y)) - RF_y * h \\
 \therefore \\
 d_y &= \frac{RF_y * h - RF_z * w}{(mg - RF_z)}
 \end{aligned}
 \tag{eq. 4.7}$$

In the above equations d_y is the distance between the COP and the COM in the global Y direction, RF_y is the reaction force vector at the hand in the global Y direction, h is the height of the point of force application at the hand relative to the floor surface, m is the

participant mass, \mathbf{g} is the acceleration due to gravity, \mathbf{RF}_z is the reaction force vector at the hand in the global Z direction, \mathbf{w} is the horizontal distance between the COM and the point of force application. Identical calculations determined the COP location in the global X direction, based on the reaction forces in the Z and X directions. The resulting COP location was expressed relative to the whole body COM (d_x, d_y), and then translated into the global coordinate system. The COP location in the global coordinate system (for each increment of hand force) was then passed on to the Maximum Force Calculator.

4.3.4 – Maximum force calculator and Monte Carlo simulation

Population variability was modeled by probabilistically selecting constraint thresholds and comparing them against outputs from the linked segment model (Figure 4.9).

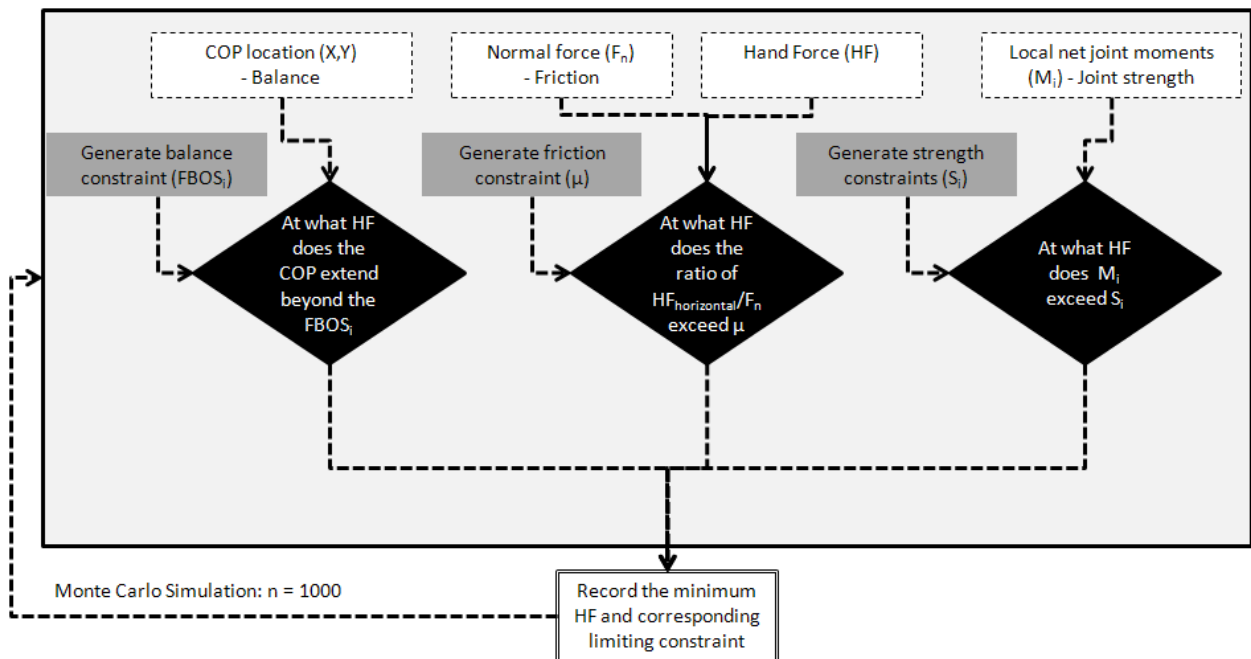


Figure 4.9 – A schematic representation of the Maximum Force Calculator.

Each linked segment model output variable was expressed as a 1000 row matrix, depicting the variable magnitude corresponding to each increment of possible hand force from 1 to 1000 N. Within the Maximum Force Calculator, each matrix was searched to determine at

which point the magnitude exceeded the probabilistically generated constraint. A Monte Carlo simulation was used to achieve this, where constraints were randomly selected for each iteration of the simulation. The process for setting and generating constraint limits within the simulation is described as follows:

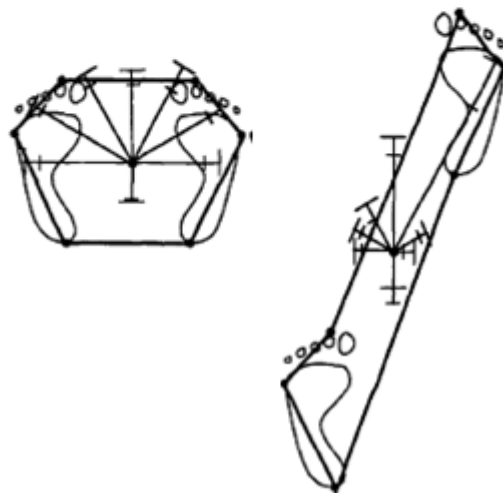
4.3.4.1 – Balance constraint

Whole body balance was implemented as a constraint by determining when the COP extended beyond the area of the base of support. However, several researchers have demonstrated that participants prefer to maintain their COP within a smaller “functional” region within the BOS termed the Functional BOS or FBOS (Holbein and Chaffin, 1997; Holbein and Redfern, 1997; Kerk et al., 1998; Lee and Lee, 2003; Holbein-Jenny et al., 2007) location with FBOS (Holbien & Chaffin, 1997; Holbein-Jenny et al., 2007). The FBOS limits are directionally defined as a percentage of the distance between the center of the BOS and the edge of the BOS. Therefore it was considered important to incorporate this level of control in the balance constraint within the model. Rather than restrict the COP to the entire area of the BOS, the COP was restricted to the FBOS limits defined in previous literature (Holbein and Chaffin, 1997). The FBOS percentages (FBOS%) used to determine the FBOS constraint (Table 1) were presented by Holbein and Chaffin (1997), and are the only description of FBOS limits for both symmetrical and asymmetrical foot placement. Since asymmetrical foot placements are often advocated as a useful technique to improve force exertion capacity (Daams, 1993), it was important to include an asymmetrical representation of balance during these types of exertions. Asymmetry was assumed when a heel marker was located in front of the 5th metatarsal head marker for the contralateral foot.

The BOS was defined in the 3DHFPM from the posture data input, based on markers placed over the lateral malleolus, the tip of the 1st metatarsal, the head of the 5th metatarsal the posterior edge of the calcaneus (heel). These landmarks were used to define the BOS, calculate the center of the BOS, and determine the length (BOS_{length}) from the BOS center to the BOS edge in the forward, backward, left and right directions. The FBOS% reported by Holbein and Chaffin (1997) were based on a three-point description for the BOS (highlighted in table 4.1). The 3DHFMP used an adjusted approach to define the BOS using a four-point estimate, shifting the 2nd tarsal marker the tip of the 1st metatarsal and adding a marker on the lateral malleolus to more accurately define the BOS (Holbein-Jenny et al., 2007). The adjustment increased the area defined as the BOS compared to the less detailed BOS definition (Holbein and Chaffin, 1997). As illustrated in Table 4.1, some limits exceed the BOS as defined by the marker data. The marker locations underestimate the actual BOS as they do not account for the additional area between the head of the metatarsals to the end of the shoe. Therefore, the COP may actual extend beyond the edge of the BOS defined as the head of the metatarsals, into the phalanges and toe of the shoe.

Table 4.1 – Functional base of support as a percentage of the base of support (bolded) and standard deviations (*italicized*) used to constrain balance in the 3DHFP. The BOS diagrams below provide a graphical representation of the FBOS limits (Holbein and Chaffin, 1997; *Reproduced with permission from Human Factors. 39(3), 456-468. Copyright 1997 by the Human Factors and ergonomics Society. All rights reserved*). The values listed in the table for both symmetric and asymmetric foot placements represent an average of wide and narrow foot placements.

FBOS (%)	Symmetric	Asymmetric
Forward	112.5	147.5
±	<i>20.5</i>	<i>33.1</i>
Backward	45.5	42
±	<i>20.5</i>	<i>30.0</i>
Right	85	54
±	<i>19.1</i>	<i>11.0</i>
Left	95.5	122.5
±	<i>14.1</i>	<i>43.0</i>



The FBOS constraint (% of FBOS length) in each direction (i) was calculated at each increment of the Monte Carlo simulation as follows:

$$FBOS_i = BOS_{length_i} * (FBOS\% + FBOS\%_{SD} * R) \quad (\text{eq. 4.8})$$

where R is a random number selected from a normal distribution with a mean of 0 and a standard deviation of 1. This approach allowed the constraint to be drawn probabilistically from a normal distribution of possible FBOS limits based on mean (FBOS%) and corresponding standard deviation (FBOS%_{SD}).

4.3.4.2 – Friction constraint

Shoe-floor friction was implemented as a constraint by comparing the ratio between horizontal hand force and normal force to a probabilistically generated coefficient of friction (μ). This relationship assumes that the maximum amount of horizontal force that can be generated at the handle is limited to the amount of frictional force available at the feet, assuming no other body-environment contact points. Further, the frictional force available is dependent on the normal force and the static coefficient of friction at the footwear-floor interface. This approach has been used previously in deterministic models to predict maximum horizontal hand forces (Grieve, 1979a; 1979b; Kerk et al., 1994). To model the available friction using a probabilistic approach the variability in static coefficients of friction was modeled based on possible coefficients depending on the type of footwear and surface. This approach to induce variability was sought, as there is no population variability in the coefficient of friction. The variability in this measure is related to the footwear and floor surfaces. To generate a distribution for μ , a mean (\bar{X}_μ) and standard deviation (SD_μ) were calculated from previously reported experimental data measured for dry leather soles over twelve different floor surfaces (Kroemer, 1974). The resulting \bar{X}_μ and SD_μ used were 0.51 ± 0.16 .

Similar to equation 8, the μ constraint was probabilistically determined for each increment of the simulation as follows:

$$\mu = \bar{X}_\mu + SD_\mu * R \quad (\text{eq. 4.9})$$

4.3.4.3 – Strength constraint

Maximum net moments at each joint were also used as constraints. Local joint moments output from the linked segment model were compared to population joint strength

limits. The mean population joint strength capability (\bar{S}_i) in Newton meters was predicted using gender specific regression equations (Schanne, 1972; Stobbe, 1982; adapted from Kumar, 1996). The regression equations estimated \bar{S}_i based on the input of joint angles. The joint angles used in the equations were described in section 4.3.2.4. The standard deviation (SD_{Si}) in joint strength about the mean was then calculated by multiplying the \bar{S}_i by the coefficient of variation reported in the literature (Stobbe, 1982).

Strength data for the elbow (flexion/extension) and shoulder (flexion/extension, ab/adduction, internal/external rotation) were incorporated directly using the regression equations of Schanne (1972) and scaled using the data presented by Stobbe (1982), as summarized in Chaffin et al., (2005). Ideally, all strength estimates would be taken from the same source to ensure that all the data was collected in the same manner. However, the Schanne (1972) dataset did not provide information on trunk rotation strength. Therefore the data provided by Kumar (1996) was used as it provided strength estimates about all three anatomical axes (flexion/extension, lateral bend, and axial twist). Kumar (1996) reported strength estimates over a series of discrete angles so regression equations were derived using a 2nd order polynomial fit (similar to the approach used by Schanne, 1972) in order to predict strength over a continuous range of angles. The α , β , and γ angles were calculated as described in section 4.3.2.4:

Extension strength (\bar{S}_{ext}):

$$\begin{aligned}\bar{S}_{ext} &= -0.0562 \alpha_{trunk}^2 + 18.475 \alpha_{trunk} - 1240 \\ R^2 &= 0.9571\end{aligned}\tag{eq. 4.10}$$

Flexion strength (\bar{S}_{flex}):

$$\begin{aligned}\bar{S}_{flex} &= -0.025 \alpha_{trunk}^2 + 8.84 \alpha_{trunk} - 619 \\ R^2 &= 0.9998\end{aligned}\tag{eq. 4.11}$$

Lateral bend strength (\bar{S}_{bend}):

$$\begin{aligned}\bar{S}_{bend} &= -0.0505 \beta_{trunk}^2 - 0.1 \beta_{trunk} + 139.05 \\ R^2 &= 0.9789\end{aligned}\tag{eq. 4.12}$$

Axial twist strength (\bar{S}_{twist}):

$$\begin{aligned}\bar{S}_{twist} &= -0.024 \gamma_{trunk}^2 + 10.0467 \gamma_{trunk} + 65.775 \\ R^2 &= 0.9696\end{aligned}\tag{eq. 4.13}$$

A coefficient of variation (CV) factor was also calculated from the trunk strength data (Kumar, 1996), similar to the approach previously used by Stobbe (1982), yielding the standard deviation (SD_{S_i}) in mean joint strengths. The CV factor was calculated for each \bar{S}_i , by dividing the standard deviation in strength by the mean measured strength at each discrete test interval. The resulting CVs were averaged across the discrete intervals to generate a representative CV. The corresponding representative CVs used in the model were: Extension strength = 0.3962; flexion strength = 0.3267; lateral bend strength = 0.2876; and axial twist strength = 0.3446.

Similar to equations 8 and 9, the joint specific strength constraint (S_i) was probabilistically determined for each increment as follows:

$$S_i = \bar{S}_i + SD_{S_i} * R\tag{eq. 4.14}$$

4.4 – Model output

A maximum force and corresponding limiting constraint is generated at each increment of the Monte Carlo simulation. At the conclusion of the simulations a mean and standard deviation of the resulting maximum forces is determined. A limiting factor frequency is also calculated by counting how many times each constraint is limiting, and then expressing that count as a percentage of the number of simulation iterations. The resulting percentage provides an indication of the probability that the designated factor is indeed limiting the exertion. The user interface shown in Figure 4.1 is presented again in Figure 4.10, now including the outputs predicted for a pulling exertion.

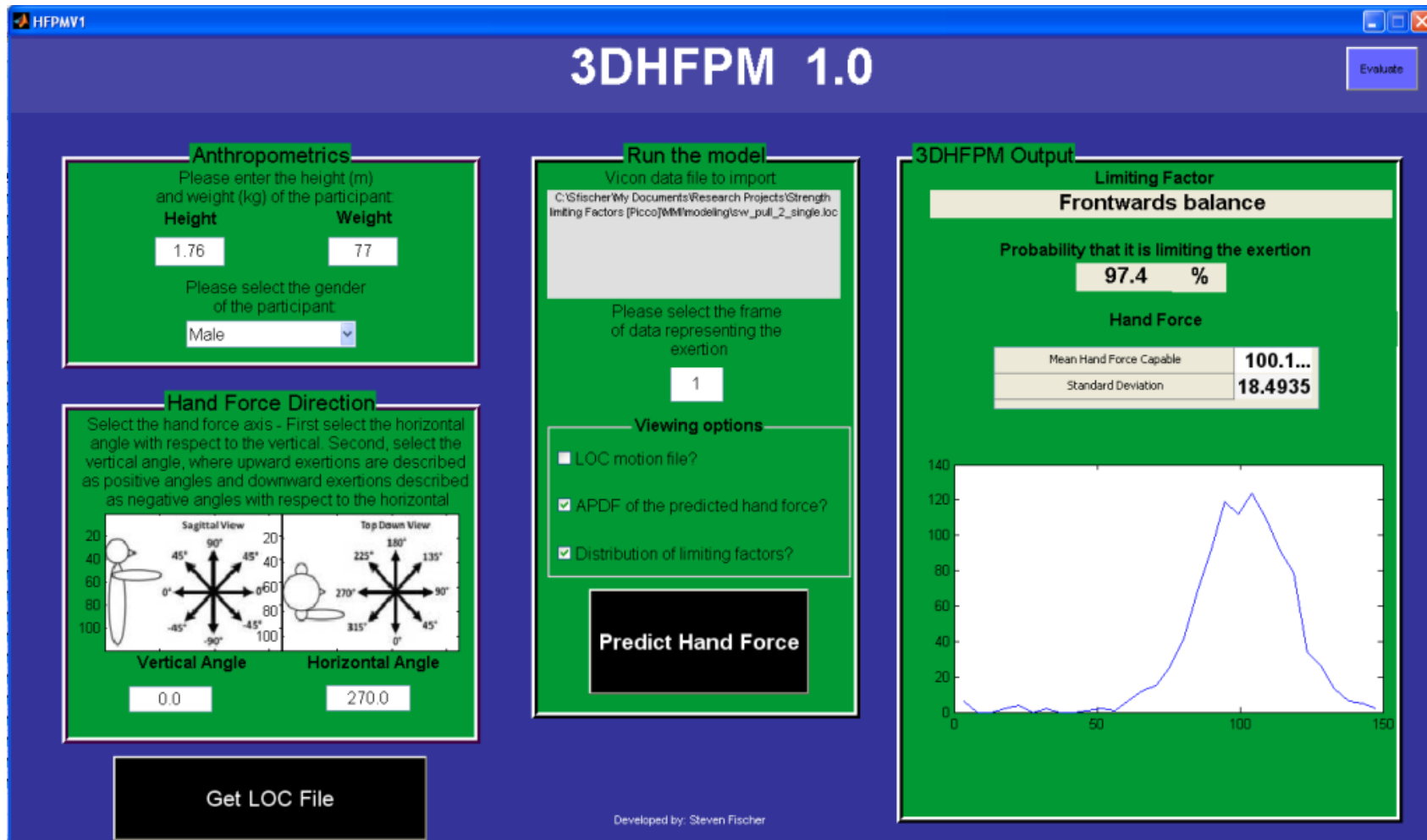


Figure 4.10 – The user interface for the 3DHFPM. Input data for a pull exertion includes: participant stature and body mass, information on hand force direction, and a posture (LOC file format currently). Output data includes the description of the limiting factor, the probability that it is the limiting factor (the frequency that it is selected as the constraint), the maximum hand force capability (N), the corresponding standard deviation, and a graph expressing the frequency (y-axis) of selecting each force level (x-axis) during the simulation.

The interface also allows a user to graph the simulated forces as a cumulative density function (Figure 4.11) and to view a histogram demonstrating the frequency for each constraint that was selected as a limiter for the described exertion (Figure 4.12).

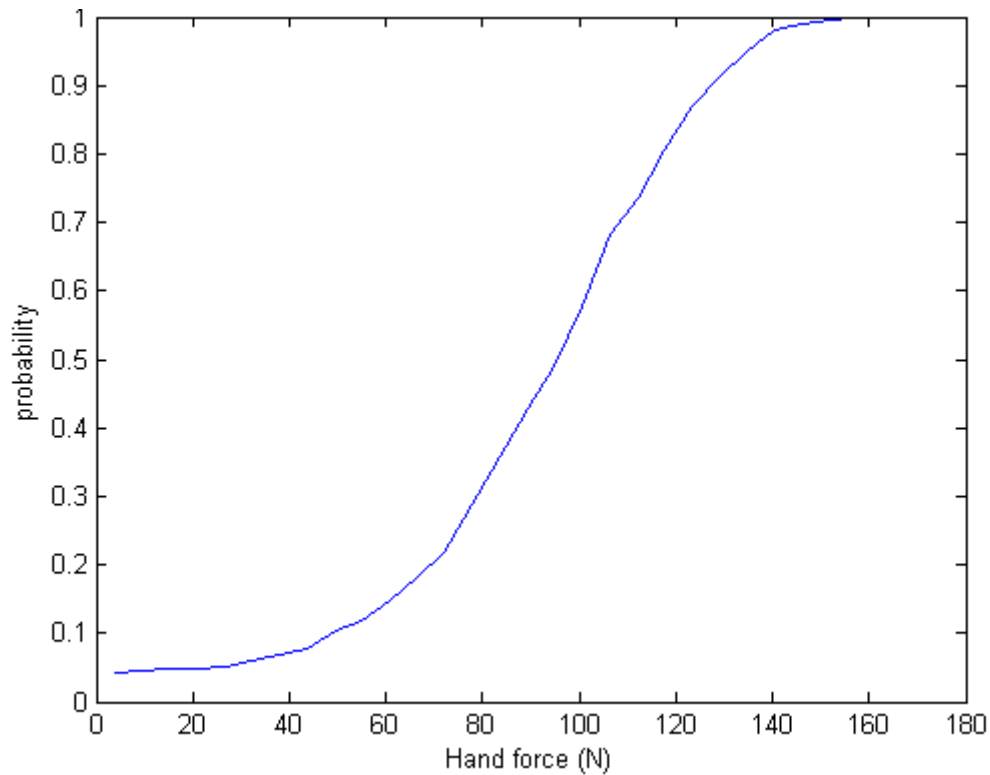


Figure 4.11 – The cumulative density function (CDF) of the maximum hand forces generated during the simulation of a medial pushing task. Assuming that the probabilistic approach provides an adequate representation of population variability, the CDF can be interpreted as the force capability across percentiles of the population.

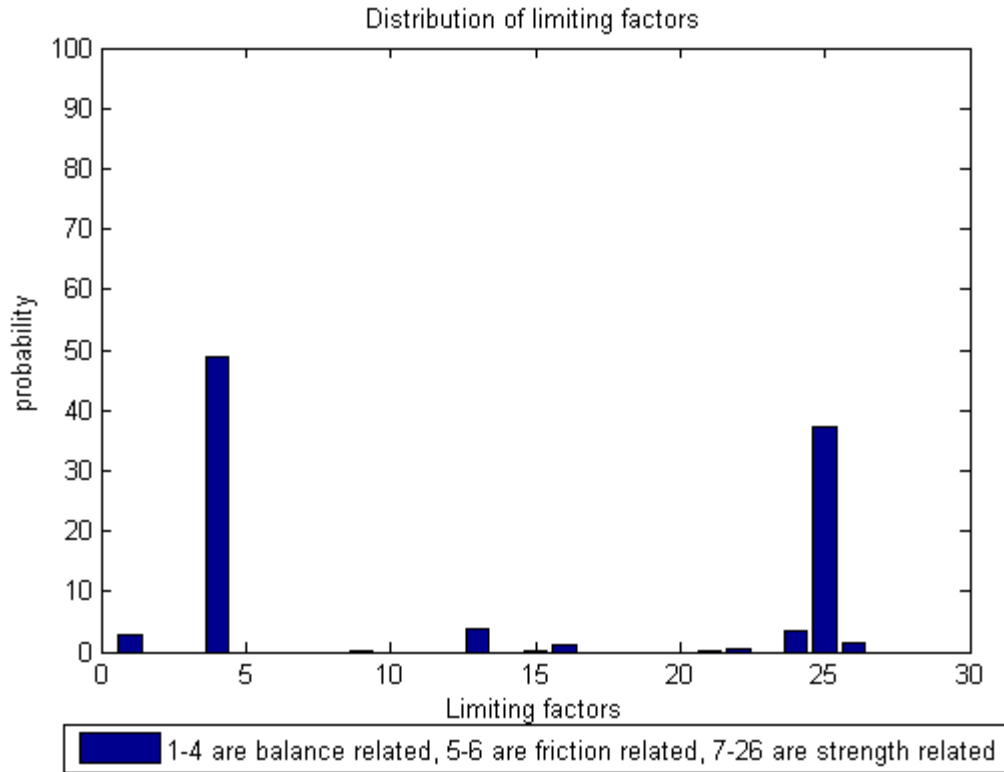


Figure 4.12 – A histogram demonstrating the probability that each constraint is a limiting factor during the simulation of a medial pushing task. The most probable limiting factor was right side balance (shown as 4 in the histogram). The second most probable limiting factor was trunk twisting strength (shown as 25 in the histogram).

The 3DHFPM provides an approach to predict both the biomechanical weakest link and the maximum voluntary hand force. The following Chapter – Part 2: model evaluation, provides an evaluation to demonstrate how well the 3DHFPM predicts these variable of interest.

Chapter 5 – A stochastic 3D static hand force prediction model for estimating maximum feasible hand forces during unilateral exertions: Part 2 – Model evaluation

Steven L Fischer, Clark R Dickerson and Richard P Wells

5.1 – Overview

In the previous chapter a model to predict maximum hand force capacity was described. This chapter presents and interprets the results of an evaluation of that model, comparing model predictions with empirically measured maximum hand force capacities. The evaluation was used to examine how well the model predicted a biomechanical weakest link, and to assess the magnitude similarity between model predicted and empirically derived maximum hand forces. Additional sensitivity analyses were performed to assess the impact of including off-axis hand forces, and to determine the impact of using a top-down approach to estimate the center of pressure location.

The model correctly identified limiting factors that matched those uncovered in the empirical investigation, 83% of the time. The model typically underestimated measured force production by 41%, 18% and 25% during downward pressing, medially pushing and horizontal pulling respectively. These predictions assumed all force was directed along the prescribed exertion direction. The underestimation was reduced to 26% during downward exertions and was not changed in the other directions when off-axis forces were accounted for. The model, when accounting for off-axis force provides a robust approach for population based force predictions.

5.2 – Introduction

Accurate estimation of occupational performance capacity facilitates better proactive job design and reactive redesign, by informing appropriate changes to job demands. In Part 1 a model to predict hand force capability was described. The model was developed using a weakest link strategy, where the maximum force capacity was governed by an underlying biomechanical factor such as joint strength (moment), shoe-floor friction, or whole body balance. In Part 2 the evaluation process and results are presented along with a general discussion of the predictive capacity of the model.

The predictive capacity of the model was evaluated by comparing experimentally measured maximum hand forces with model predicted hand forces. The experimental data was collected as part of a research project investigating the role of whole-body balance, shoe-floor friction and joint strength on hand force capability (Chapter 3). An overview of the methodology is provided along with specific details describing how the motion data was used to drive the model predictions.

5.3 – Methods

5.3.1 – Experimental data

Hand force and posture data were collected while participants exerted maximal forces in a series of constrained and unconstrained positions. Participants completed two repetitions of a horizontal pull, a medial push, and a downward press in each of five experimental conditions (Figure 5.1): shoulder width foot placement (SWFP), free foot placement (FFP), high friction – feet shoulder width (HF – to reduce the influence of friction as a limiter), lower body braced (LBB – to eliminate the influence of both friction and balance as force limiters),

and both the lower and upper body braced (UBB – to eliminate the influence of both balance, friction and torso strength).

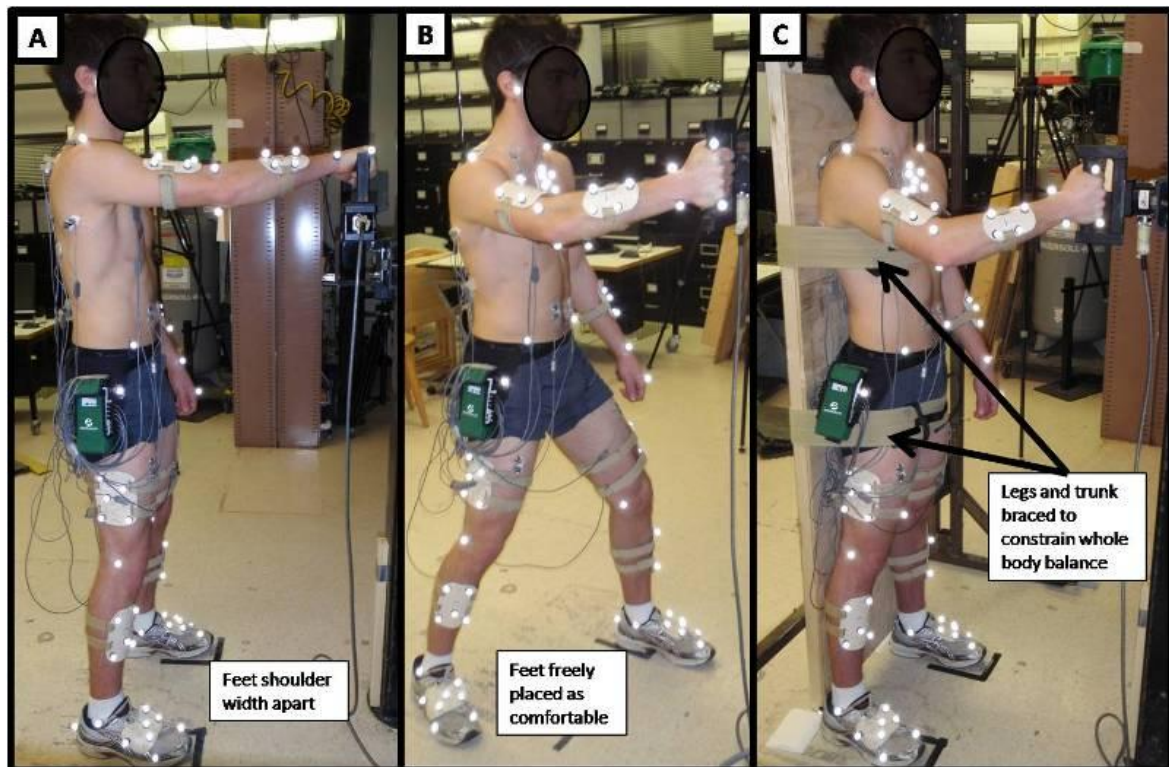


Figure 5.1 – The experimental conditions tested. Participants completed downward presses, pulls, and medial pushes in five conditions: 1) shoulder width foot placement (“SWFP” seen in frame A), 2) free foot placement (“FFP” seen in frame B), 3) high friction (“HF” seen in frame A, with feet taped to floor), 4) lower body braced (“LBB” seen in frame C, without upper body chest strap), and upper body braced (“UBB” seen in frame C).

The postural data input into the model was derived from the experimental marker data. Posture data was described in a pre-defined format, created from the experimental marker data. The predefined format was selected to allow the same data to also be used with the Michigan three-dimensional static strength prediction program (3DSSPP) and the Shoulder Loading Analysis Modules (SLAM) program (Dickerson et al., 2007). The use of the format was deliberate to allow experimental data to be easily used in other biomechanically based models to facilitate comparison where appropriate. The method used for extracting joint centers from the experimental data was described in Part 1. The specific joint centers and experimental

markers used to describe posture are shown in Figure 5.2 and the abbreviations are presented in Table 5.1.

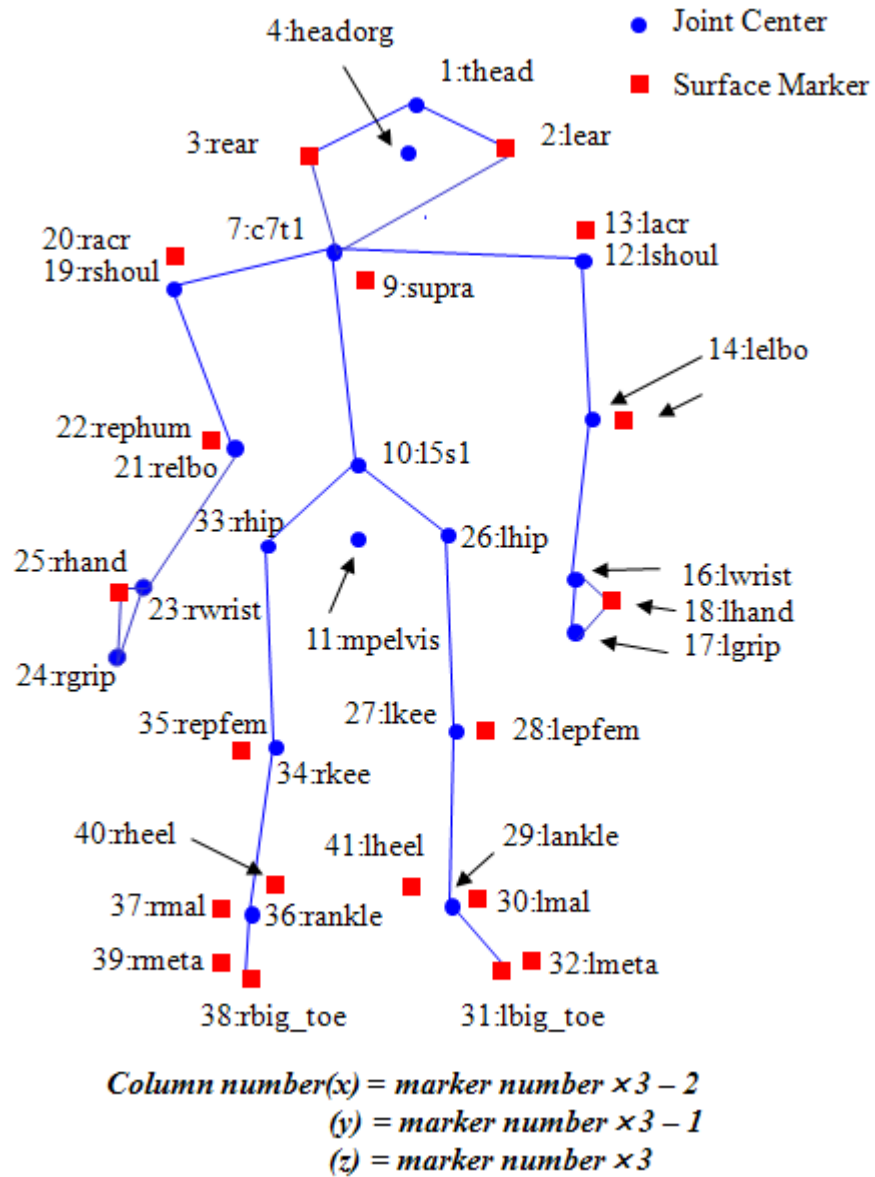


Figure 5.2 – An illustration of the joint center and surface marker data used to describe the whole body posture to be inputted into the model. Note that not all columns in a particular file will have numeric data (i.e. they may be zeros), depending on the collection.

Table 5.1 – A description of the abbreviations used to describe the marker locations in Figure 5.2.

Marker number	Abbreviation	Name	Marker number	Abbreviation	Name
1		unused	26	lhip	Left hip joint center
2	lear	Left ear	27	lknee	Left knee joint center
3	rear	Right ear	28	lepfem	Left femoral condyle
4	headorg	Head COM	29	lankle	Left ankle joint center
5		unused	30	lmal	Left lateral malleolus
6		unused	31	lbig_toe	Left 1 st distal phalange
7	c7t1	C7 vertebra	32	lmeta	Left 5 th metatarsal head
8		unused	33	rhip	Right hip joint center
9	supra	Suprasternal notch	34	rknee	Right knee joint center
10	l5s1	L5 vertebra	35	repfem	Right femoral condyle
11	mpelvis	Center of the pelvis	36	rankle	Right ankle joint center
12	lshoul	Left humeral head center	37	rml	Right lateral malleolus
13	lacr	Left acromion	38	rbig_toe	Right 1st distal phalange
14	lelbo	Left elbow joint center	39	rmeta	Right 5th metatarsal head
15	lephum	Left lateral epicondyle	40	rheel	Right heel
16	lwrist	Left wrist joint center	41	lheel	Left heel
17	lgrip	Left grip center			
18	lhand	Left hand center			
19	rshoul	Right humeral head center			
20	racr	Right acromion			
21	relbo	Right elbow joint center			
22	rephum	Right lateral epicondyle			
23	rwrist	Right wrist joint center			
24	rgrip	Right grip center			
25	rhand	Right hand center			

5.3.2 – Evaluation criteria

The model was evaluated on two criteria: identification of the limiting constraint (content validity), and magnitude similarity (criterion validity). Evaluation of the accuracy of constraint identification was performed as the percentage of correctly identified constraints (where constraints were identified experimentally and described in Chapter 3). Magnitude similarity was quantitatively assessed using repeated measures ANOVA and Spearman Rho correlations within each force direction.

The experimental conditions were chosen to be limited by different factors through the use of different foot placements and constraints. Therefore, when modeling the force output the model was also changed to systematically remove constraints matching the experimental conditions. All constraints were enabled when modeling the data from the SWFP and FFP experimental conditions. The friction constraint was disabled in the model during the HF condition. Both balance and friction were disabled in the model while predicting hand force for the LBB and UBB conditions and trunk strength was also disabled during the UBB conditions. The systematic removal of model constraints, corresponding to experimental conditions where constraints were removed, helped evaluate the feasibility of using a weakest link strategy. Theoretically, if the weakest link is removed experimentally and hand force increases, the model should predict a similar increase in force when the same constraint is disabled in the model.

Two additional tests were performed during the evaluation process: an off-axis force test, and a COP calculation test to determine the effects of the underlying assumptions related to these calculations. Off-axis force production during any maximum exertion can alter the net force direction from a prescribed direction (Abel et al., 1991; Fothergill et al., 1991; de Looze, et al., 2000; Granata and Bennett, 2005; Hoozemans et al., 2007; Hoffman et al., 2007). The model does not account for this alteration and assumes the force is exerted in the direction prescribed by the user's input. To determine the effect of this assumption, the model was evaluated in its current state (assuming no change in the force vector direction), and again by revising the prescribed direction to match the actual direction. A unit vector describing the actual force direction was determined from the components of the experimentally measured force data (See Figure 4.7).

The model described in Part 1 relied on the center of pressure (COP) being estimated, rather than measured using force plate data. An “impact of COP assumptions” evaluation was conducted to compare the 3DHFPM COP estimates with those obtained from a force plate. One male participant completed two repetitions of a pull, a medial push and a downward press in the SWFP condition, while standing on a force plate. The COP calculated from the force plate was compared to the COP calculated from the 3DHFPM to determine any differences.

5.3.3 – Statistical analysis

A one-factor repeated measures ANOVA was used to detect for significant differences between the measured, model predicted (no-off axis) and model predicted (off-axis included) hand force in each of the fifteen different experimental conditions to determine if the model(s) significantly under- or over predicted measured values. The measured value and two model predicted values represented the repeated measure within each participant, within each of the fifteen conditions. A Greenhouse-Geisser correction was used to correct the degrees of freedom when the data violated the sphericity assumption. Pairwise comparisons were used post hoc to detect the location of the significant differences. A repeated measures ANOVA was selected for two main reasons: The repeated measures design accounts for the correlation between the measures, as the modeled and measured hand forces are not independent, violating the assumption of the ANOVA, and second, the repeated measures helps to account for the large inter-participant variability in hand force capability. The Spearman Rho was used to determine if the model readily predicted higher forces for individuals producing higher hand forces. Rank ordered correlations were calculated between measured and predicted hand force means in each of the fifteen experimental conditions (n=18). The Spearman Rho was applied to detect if the model predicted higher hand forces in conditions where participants also

produced higher hand forces. The Spearman Rho was selected because of the limited sample size. A larger sample would be needed to ensure a normal distribution, which is required to support a parametric alternative Alpha was set at 0.05 for all statistical comparisons. All statistical analysis was completed using SPSS (SPSS INC., Chicago, IL, USA).

5.4 – Results

5.4.1 – Constraint selection

The most frequently predicted limiting factor for each condition correctly matched the experimentally determined limiting factor for ten of the fifteen conditions (Table 5.2). Mismatches occurred twice; in the SWFP and FFP conditions during downward pressing, where the model selected backwards balance as limiting while the biomechanical data pointed to shoulder or elbow strength as limiting. For the remaining three conditions (medial SWFP, FFP and HF), an experimentally determined limiting factor was unidentifiable within that study's design, as it may have been related to a rotational balance constraint that was not controlled or tested for. Therefore there was no experimental data to compare model estimates against for these conditions.

Table 5.2 – Comparison of experimentally determined and model predicted limiting factors by experimental scenario. Agreements on the limiter are shown in bold and *italicized*.

Condition	Downward exertions			Medial exertions			Pulling exertions		
	Experimental Limiter	Model Predicted Limiter	Limiter Probability	Experimental Limiter	Model Predicted Limiter	Limiter Probability	Experimental Limiter	Model Predicted Limiter	Limiter Probability
SWFP	Shoulder or Elbow Strength	Backward balance	69%	Unclear	Trunk twist strength	60%	<i>Frontward Balance</i>	<i>Frontward balance</i>	95%
FFP	Shoulder or Elbow Strength	Backward balance	63%	Unclear	Trunk twist strength	54%	<i>Frontward Balance</i>	<i>Frontward balance</i>	91%
HF	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	71%	Unclear	Right side balance	64%	<i>Frontward Balance</i>	<i>Frontward balance</i>	93%
LBB	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	80%	<i>Shoulder or trunk strength</i>	<i>Trunk twist strength</i>	73%	<i>Elbow or trunk strength</i>	<i>Trunk extensor strength</i>	56%
UBB	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	86%	<i>Shoulder or trunk strength</i>	<i>Shoulder internal rotation strength</i>	62%	<i>Elbow or trunk strength</i>	<i>Elbow flexor strength</i>	64%

5.4.2 – *Magnitude similarity*

The model predictions underestimated experimental hand forces by an average of 18, 26, and 41% during medial, pulling and downward exertions respectively (Table 5.3). Model predictions were not different from experimentally measured forces (Figure 5.3) in three experimental conditions (medial exertions in the SWFP and UBB conditions, and when pulling in the UBB condition). Despite these underestimates, the Spearman Rho statistic demonstrated that the rank ordered predicted values were significantly associated with the rank ordered measured force values in twelve of the fifteen exertion scenarios (Table 5.4). No significant associations were seen during downward exertions in the FFP and UBB conditions, or during pulling in the FFP condition. The lack of a significant association in the FFP conditions, and the demonstrated relative differences observable between model and predicted hand forces in the FFP condition shown in Figure 5.3 may provide evidence that the current approach for constraining balance during asymmetric exertions (FFP) may be overly restrictive.

Table 5.3 – Results from the Repeated Measures ANOVA statistical model. Repeated measures were compared between the measured hand force, and two model predicted hand forces (with and without accounting for off-axis forces). Both models adequately predicted measured forces during medial exertions in the SWFP and UBB conditions.

Condition		df	F	p	Partial Eta Squared	Observed Power
SWFP - Down	Sphericity Assumed	2	93.651	<0.001	.824	1.000
SWFP - Medial	Greenhouse-Geisser	1.165	1.278	.277	.060	.202
SWFP - Pull	Greenhouse-Geisser	1.044	15.854	.001	.442	.971
FFP - Down	Sphericity Assumed	2	63.973	<0.001	.762	1.000
FFP - Medial	Greenhouse-Geisser	1.344	16.895	<0.001	.458	.993
FFP - Pull	Greenhouse-Geisser	1.008	32.529	<0.001	.619	1.000
HF - Down	Greenhouse-Geisser	1.354	51.547	<0.001	.720	1.000
HF - Medial	Greenhouse-Geisser	1.017	13.383	.001	.401	.938
HF - Pull	Greenhouse-Geisser	1.138	22.980	<0.001	.535	.998
LBB - Down	Greenhouse-Geisser	1.325	50.105	<0.001	.715	1.000
LBB - Medial	Greenhouse-Geisser	1.164	20.968	<0.001	.512	.996
LBB - Pull	Greenhouse-Geisser	1.115	14.129	.001	.414	.962
UBB - Down	Greenhouse-Geisser	1.306	28.428	<0.001	.587	1.000
UBB - Medial	Greenhouse-Geisser	1.222	2.728	.105	.120	.388
UBB - Pull	Greenhouse-Geisser	1.554	22.818	<0.001	.533	1.000

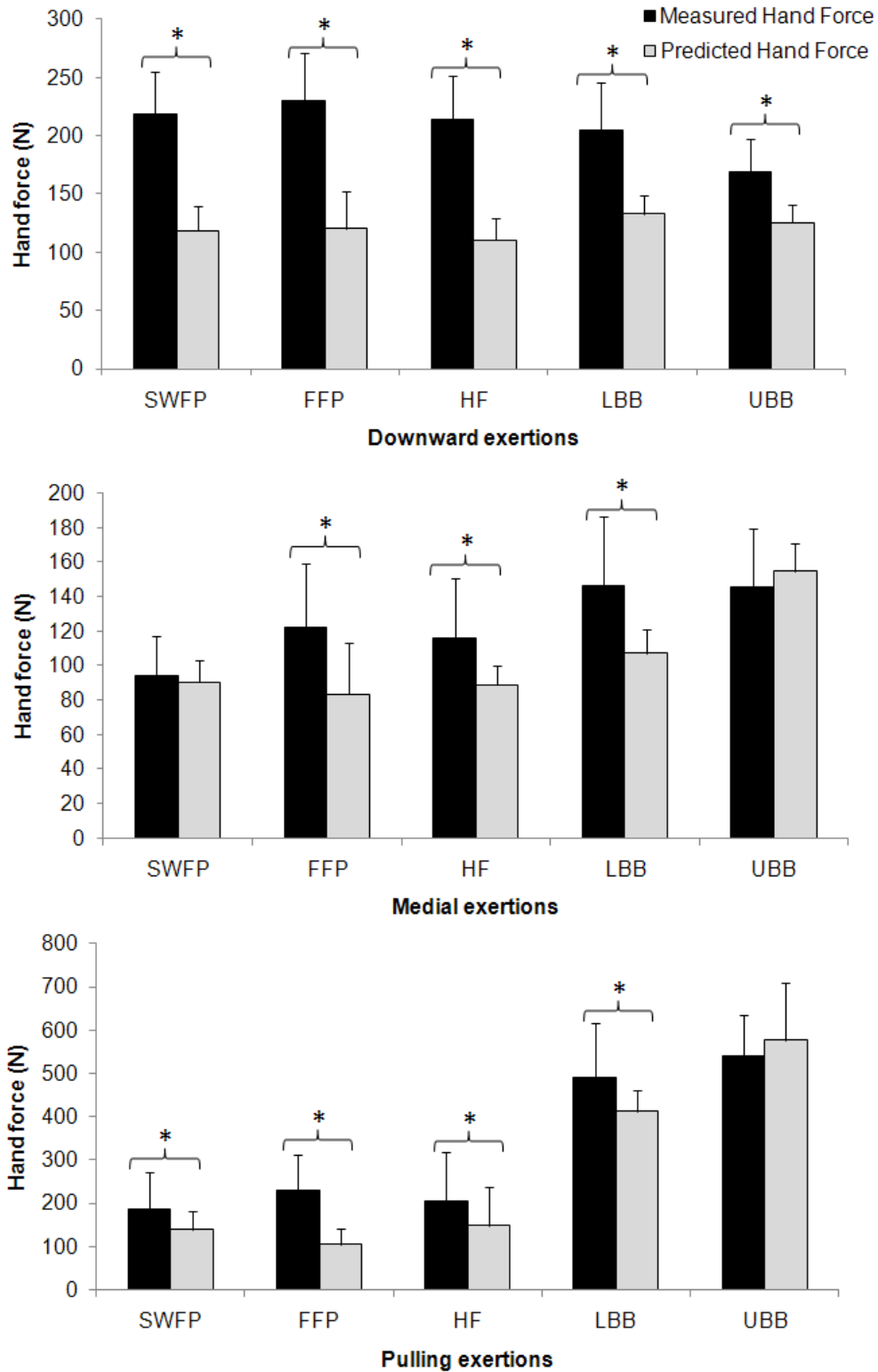


Figure 5.3 – Measured and predicted hand forces during fifteen different exertions scenarios. Significant pairwise differences ($p < 0.05$) between measured and predicted hand forces (no off-axis forces) are denoted with an asterisk (*).

Table 5.4 – Spearman Rho correlation coefficients between measured and modeled hand forces. Significant associations are bolded ($p < 0.05$).

Downward exertions					
Condition	SWFP	FFP	HF	LBB	UBB
R	0.506	0.202	0.411	0.505	0.246
p-value	0.120	0.344	0.046	0.012	0.246
Medial exertions					
Condition	SWFP	FFP	HF	LBB	UBB
R	0.702	0.436	0.905	0.543	0.544
p-value	0.000	0.033	0.000	0.006	0.006
Pulling exertions					
Condition	SWFP	FFP	HF	LBB	UBB
R	0.930	0.186	0.918	0.551	0.427
p-value	0.000	0.385	0.000	0.005	0.053

5.4.3 – Impact of including off-axis forces

The use of the actual force vector improved the capacity of the model to identify limiting constraints. The number of correctly identified constraints improved from ten to eleven of the fifteen conditions tested (Table 5.5).

The use of the actual force vector also improved the magnitude similarity during all downward exertion scenarios, and during pulling in the SWFP scenario (Figure 5.5). For downward exertions, the revised approach using the actual force vectors improved the underestimate from 41% to 26% on average across conditions. During pulling in the SWFP condition, including the influence of off-axis forces by incorporating the actual force vector significantly improved estimates by a more modest 4%. Using the actual force vector had no significant impact on predictions for medial exertions, and the model therefore continued to underestimate hand force during these exertions in the FFP, HF, and LBB scenarios.

Table 5.5 – Comparison of experimentally determined and model predicted limiting factors by experimental scenario, incorporating off-axis forces. Correct matches are shown in bold and *italicized*.

Condition	Downward exertions			Medial exertions			Pulling exertions		
	Experimental Limiter	Model Predicted Limiter	Limiter Probability	Experimental Limiter	Model Predicted Limiter	Limiter Probability	Experimental Limiter	Model Predicted Limiter	Limiter Probability
SWFP	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	57%	Unclear	Trunk twist strength	61%	<i>Frontward Balance</i>	<i>Frontward balance</i>	85%
FFP	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	53%	Unclear	Trunk twist strength	56%	<i>Frontward Balance</i>	<i>Frontward balance</i>	83%
HF	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	63%	Unclear	Right side balance	63%	<i>Frontward Balance</i>	<i>Frontward balance</i>	82%
LBB	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	67%	<i>Shoulder or trunk strength</i>	<i>Trunk twist strength</i>	68%	<i>Elbow or trunk strength</i>	<i>Trunk extensor strength</i>	70%
UBB	<i>Shoulder or Elbow Strength</i>	<i>Right elbow extensor strength</i>	76%	<i>Shoulder or trunk strength</i>	<i>Shoulder internal rotation strength</i>	64%	Elbow or trunk strength	Shoulder extensor strength	53%

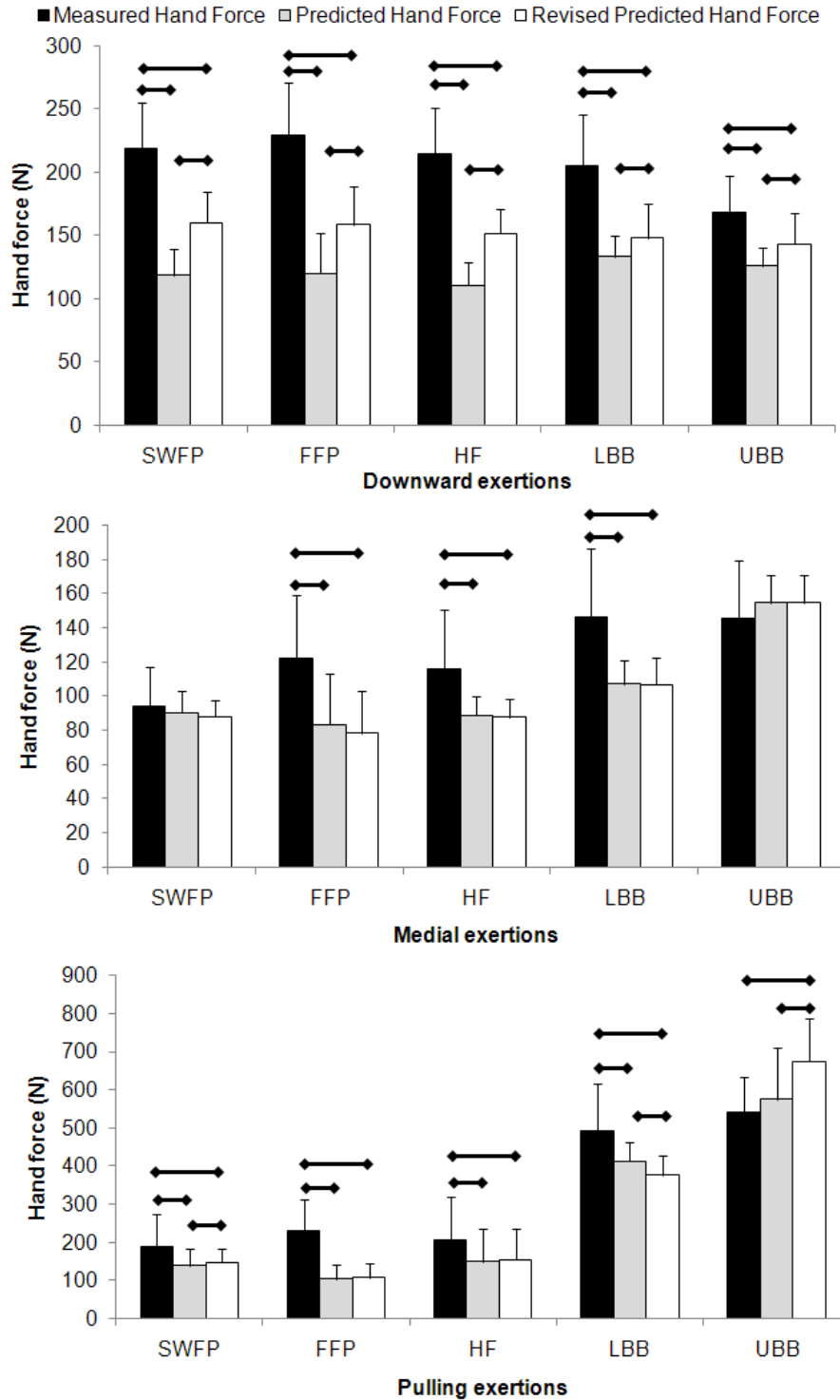


Figure 5.4 – Measured, predicted and revised (inclusion of off-axis force) predicted hand forces during fifteen different exertions scenarios. Significant pairwise differences ($p < 0.05$) between measured, predicted and revised predicted hand forces are denoted with line above the significant data points.

5.4.4 – COP evaluation

The 3DHFPM overestimated the COP distance from the centre of the BOS compared to force plate estimates (Figure 5.5). During downward and medial exertions the 3DHFPM overestimated the global X (M/L) locations by 4 cm and 1.0 cm, and the global Y (A/P) locations by 2 cm and 2 cm, respectively. These differences had little impact on the model selecting the correct limiting factor as these exertions were not often limited by balance constraints. During pulling the 3DHFPM overestimated the distance of the COP from the centre of the BOS in the A/P direction by approximately 6 cm.

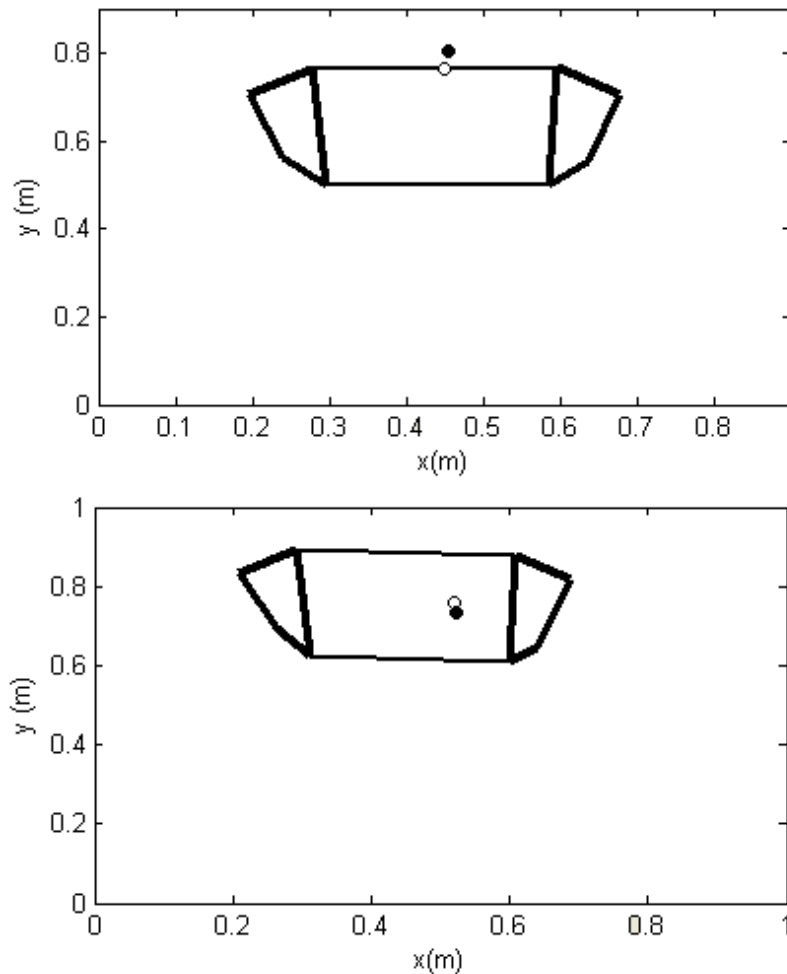


Figure 5.5 – A graphical representation of the calculation of the COP using the 3DHFPM top-down approach (filled circles) and a bottom-up force plate approach (empty circles) during pulling (top plot) and medial pushing (bottom plot).

5.5 – Discussion

The 3DHFPM was intended to facilitate the prediction of both maximum hand force capability and also to identify the factor most likely limiting that capacity, based on a weakest link approach. The evaluation demonstrated that the model does have moderate predictive capacity in that the predicted force limiters matched with theoretical force limiters in 83% of the scenarios where a theoretical limiter was identified. Further, the correlation analysis demonstrated significant associations between the rank-ordered measured and predicted hand forces, which indicates that the model typically predicted higher forces for scenarios where participants also produced higher measured forces. However, the repeated measures ANOVA indicated a significant under-prediction in twelve of the fifteen evaluated scenarios, which restricts its usefulness as a means to determine absolute maximum capability. The magnitude of under-prediction could be reduced from 41 to 26% for downward directed exertions by incorporating off axis forces, though the predictions were still less than experimental forces. The combination of a significant difference and a significant association suggests that the model responds to participant specific postures; however, the constraint values appear overly restrictive.

5.5.1 – Interpretation of the model evaluation results

The observed underestimates of maximum hand force was anticipated based on previous research. Kerk et al., (1998) also reported an underestimation when using their model to predict maximum hand force capacity during bilateral pulling and downward pressing while standing with feet positioned shoulder width apart. Similar to the evaluation results from the 3DHFPM, Kerk et al., (1998) demonstrated that forward balance was the limiting factor during pulling exertions, though they underestimated capability by 36%. During pulling in the SWFP

condition the 3DHFPM under-estimated pulling force by 25%. Though the exertions were different (bilateral versus unilateral), they shared the same limiting factor. Since the 3DHFPM used a balance constraint that is less restrictive than the one presented by Kerk et al., (1998) and still under-estimated force, it may indicate that the true forward balance limit may be further forward than the FBOS limits posed by Holbein and Chaffin (1997) and used in the 3DHFPM.

The underestimation may also be a result of the underlying data used to describe the constraint thresholds. The original strength data used to determine joint strength constraints may not be well suited for use in predicting strengths for a modern day workforce. Research has shown secular changes in a number of health and performance related measures (Tomkinson and Olds, 2007; Danubio and Sanna, 2008). Specifically in terms of strength, Westerstahl et al. (2003) reported a significant increase in isometric strength in young males from 1974 to 1995. Further, Schanne (1972) described that the strength thresholds developed using a regression model underestimate empirical strength thresholds by approximately 30%, approximately the underestimation noted in this study. This is a result of the assumption that joint strengths are independent. In the underlying regression equations (Schanne, 1972; Stobbe, 1982) moment loading at adjacent articulations does not affect the strength capability of a particular muscle group. The assumption of independence may be impacting the ability to adequately determine joint strength thresholds.

5.5.2 – Importance of off-axis forces

Including the influence of off-axis forces either did not change or improved estimate accuracy. This has two implications: it provides a rationale for supporting the inclusion of off-axis force estimates in determining maximum hand force capacity; and the influence of

including off-axis forces on the selection of limiting factors in the model provides context to current discussion in the literature attempting to explain how off-axis forces are selected to protect against possible weakest links, such as the shoulder or trunk. Since the inclusion of off-axis forces did not diminish the predictive capacity (except in the case of pulling while completely braced), and in particular, off-axis force inclusion during downward exertions significantly improved the predictive capacity, it is suggested that an approach to predict off-axis forces be implemented in future revisions of the model.

Off-axis forces are theoretically believed to help minimize joint moments.

Experimentally, the moment arm between the actual force vector and either the shoulder or L4/L5 joint, can be compared with the moment arm between the desired force vector (no off-axis forces) and the shoulder or L4/L5 joints (de Looze et al., 2000; Granata and Bennett, 2005; Hoffman et al., 2007). The concern with this approach is that the choice of joint for comparison seems to be unrelated to the mechanical weakest link. All three studies investigated pushing and pulling tasks, which have been identified as being limited by balance both within this study and previously (Kerk et al., 1998). Logically, it would be ideal to create off-axis forces that protect against a limiting factor, which in the case of horizontal pushing and pulling, is most likely balance. Therefore, it is not surprising that a clear consensus has not emerged in the literature as to whether off-axis forces are protective to trunk or shoulder joint moments.

When using the 3DHFP to predict force for pulling exertions, allowing for off-axis forces did not significantly increase hand force. However modest increases were observed with a concurrent decrease in the probability that balance was limiting (comparing the limiter probability for pulling in SWFP, FFP, and HF between Tables 5.3 and 5.5). The decrease in

probability suggests that the off-axis force has helped to protect against a balance limitation, shifting this limit to another location about 10% more often. During downward exertions, the inclusion of off-axis forces helped to protect against a backwards balance limitation, so much so, that it was no longer deemed the limiting factor. Therefore, it is likely that off-axis forces do act to minimize moments; however the moment to be minimized may be more closely related to a current exertion limiting factor, rather than strength at a specific joint.

5.5.3 – Importance of the COP calculation

Error existed between the 3DHFPM COP and the COP determined from force plate data. The 3DHFPM overestimated the COP location. This may be due in part to the errors associated with the use of anthropometric tables to estimate the whole body COM, and due to the assumption of static equilibrium. As illustrated in Figure 5.5 (top plot – a pulling example), this error may be problematic in that the 3DHFPM predicts that the COP has moved beyond the BOS limit, when the force plate based COP indicated that the COP is on the edge of the BOS limit. Therefore the 3DHFPM would likely underestimate hand force because of an overestimation of the COP location in the A/P direction. As demonstrated in the magnitude similarity section, the 3DHFPM does indeed underestimate when balance is a limiter during pulling. Future revisions of the model will aim to correct the overestimation of the COP A/P location to improve the magnitude similarity of the model to measured values when balance is limiting. Since the purpose of the model is to ultimately be used in the field, it is not ideal to require force plate information and therefore future work must aim to better understand why this mismatch exists, and then develop an appropriate corrective adjustment.

5.5.4 – Model limitations

One purpose of a model is to provide a realistic and simple approach to predict or explain phenomena. Although some of the more critical elements limiting the predictive capacity have already been discussed, several additional considerations should inform model output interpretation, including the representation of joint strength.

Joint strength capacity was predicted using strength databases derived from experimental research. The data selected for the strength constraint in the 3DHFPM (Schanne 1977; Stobbe, 1981; Kumar 1996) include the errors associated with using a regression equation to predict strength at continuous, unmeasured intervals. However, in order to detect the sensitivity of the model to these data, it may be prudent to evaluate the model using alternative sources of data. For example, Holzbaaur et al., (2007) reported a maximum elbow flexion moment of approximately 65 Nm with the elbow flexed to 90°, where the equations provided by Schanne (1977) and Stobbe (1981) would predict an elbow flexion moment of 77 Nm, an 18% increase in projected capacity. Future model revisions will be accompanied by a sensitivity analysis with regard to the impact of the strength data used to determine the joint moment constraints in the model.

Additional limiting factors may impact hand force capability. Currently the 3DHFPM incorporates joint strengths for the trunk and upper limbs, balance, and shoe-floor static friction. Recent research has identified the impact of friction at the hand-handle interface as a probable limiter for hand force capability in some situations (Greig and Wells, 2004; Seo et al., 2010). The forces and moments transmitted by the hand are sensitive to the grip type and the direction of force exertion (Greig and Wells 2004). Force transmission is also heavily influenced by friction, when the handle is oriented in parallel with the desired exertion direction (Seo et al., 2010). In the current 3DHFPM these effects are not modeled. However,

the empirical data set used in the current evaluation was generated with the hand oriented perpendicularly to the desired force direction. This provided a mechanical interference at the hand preventing hand-handle frictional properties from limiting those exertions (Seo et al., 2010).

The impact of static versus dynamic modeling approaches can be substantial (Leskinen et al., 1983). The 3DHFPD was purposefully designed for a static case despite these differences. In the current model, a user only needs postural data for a single instant in time which could be derived from photographs, where a dynamic analysis would require either estimation or measurement of the time series of motion, requiring more time and effort.

The posture data input into the model carries the assumption that it is representative of the population average postural response required to produce a maximum unilateral hand force for a person of the stature and body mass described. The posture data used to drive the model was selected at the same time point where the peak force was recorded to ensure that the posture was representative of that required to produce a maximum force. In terms of postural consistency across populations, Hoffman (2008) found that participants chose similar postural strategies when pushing or pulling; however their tactics were more varied during up/down exertions. This suggests that the postural assumption is less likely to impact the predicted population variability in pushing or pulling strength, but may be more impactful on predictions of population variability in up/down maximum unilateral hand forces.

The use of unilateral exertions for evaluation and current design was deliberate. The addition of external forces to the contra-lateral hand requires an approach to mathematically parse out the net external forces acting at each limb. Experimentally, this was achieved using separate instrumented handles, however theoretically this is more challenging to realize.

Therefore, the initial focus has been to examine unilateral exertions, allowing the focus to be on the correct identification of force limiters, and the inclusion of a probabilistic model which presents a significant advance to previous modeling approaches.

5.5.5 – Practical uses for the 3DHFFPM

The 3DHFFPM makes significant contributions to the occupational biomechanics community by providing an approach to predict both maximum hand force capability and the factor most likely limiting the exertion, or the ‘weakest link’. These predictions are important for both field and experimental applications. As a tool for field assessment, the model is robust, non-invasive, quick, and provides reasonable predictive power, meeting many of the attributes that are ideal in a field assessment tool (Hamrick, 2006). The probabilistic aspect of the model provides a significant advance by providing users with a distribution of maximum hand force capability, which is based on the population variability in strength, balance and friction. This allows users to alter task force requirements to be acceptable for a given percentage of the population. For example, design could be conducted to ensure 75% of the population could produce a specified hand force. This corresponds to the force value at 0.25 on a cumulative density function (Figure 5.6).

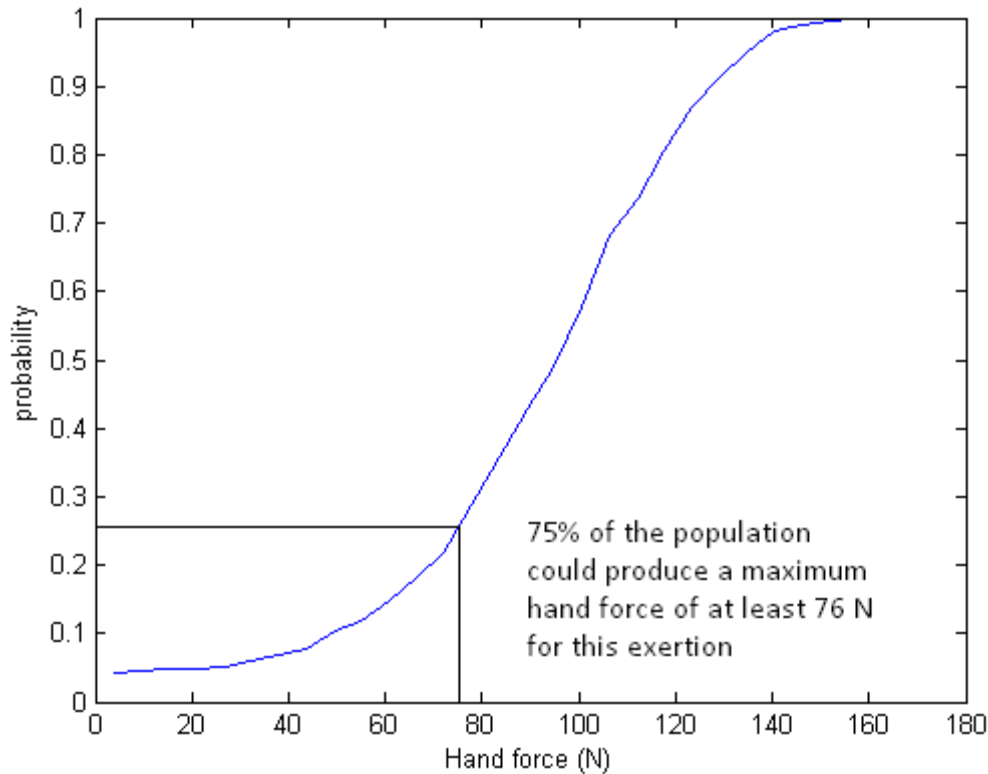


Figure 5.6 – An amplitude probability distribution function output from the 3DHFPM for a medial exertion. For the prescribed exertion the model predicted a range of possible forces with the probability of their likelihood given the population variability in the modeled constraints.

From an experimental aspect, the 3DHFPM provides the ability to understand the complex biomechanical interactions that occur during maximum force production. Specifically, the model is able to help explain the biomechanical impact of off-axis forces, on both shoulder and trunk moments, and also on balance. This level of explanation has not yet been provided in the peer reviewed literature. Additionally, further revisions to the model will help aid in understanding the influence and biomechanical rational to support bracing in the workplace, potentially as a means to protect against the ‘weakest link’ (Jones et al., 2010). Currently little information exists to explain why workers choose to brace or support themselves in the work environment while producing forceful exertions. It is likely that bracing or supporting is used to help provide resistance against a weakest link. However, the

3DHFP is well situated to help explain the biomechanical impact of bracing on the selection of a weakest link and the resultant hand force capability. The aim of future work will be to extend the predictive capacity of the 3DHFP for use during bilateral exertions better situating it as a tool to investigate the influences of bracing.

5.6 - Conclusion

The 3DHFP was developed to predict maximum hand force capacity during unilateral exertions. Further, the use of a probabilistic approach for selecting constraint limits provides a significant contribution by providing an estimate of the expected population variability about the predicted hand force capability. Lastly, the identification of limiting factors is important as their description may help identify regions of concern for potential injury, and moreover, the limiting factor may be useful in helping to predict sub-maximal limits.

The evaluation revealed that the current form of the model usually underestimated hand force capability compared to measured hand forces. However it appears that a ‘weakest link’ principle for predicting maximum force capacity is plausible, as evidenced by the significant rank ordered correlations between the measured and predicted hand forces. The 3DHFP shows promise as a possible method to estimate maximum unilateral hand force capability. However it can be improved by addressing the off-axis force concern, addressing the assumption of independence in joint strength measures and by improving the specificity of the underlying data used to establish the constraints.

Chapter 6 – Relationships between psychophysically acceptable hand forces, maximum voluntary hand force capacity and underlying biomechanical limitations

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6.1 – Overview

Psychophysical approaches are commonly used to derive thresholds for exposure to occupational demands. However, there is little biomechanical evidence to support that psychophysically derived thresholds are appropriate. This research explored the possibility that a proportional relationship exists between maximum voluntary force (MVF) capability and psychophysically acceptable forces. Additionally, we sought to determine if the magnitude of the proportionality was dependent on the biomechanical factor that limited the capacity to perform and MVF.

Seventeen male participants completed unilateral MVF exertions and determined their psychophysically acceptable force (PAF) for nine defined conditions. Proportionalities were determined by dividing the PAF by the corresponding MVF. Center of pressure and joint moments were calculated during MVF trials and used to determine if the limiting factor was balance or joint strength. The corresponding proportional values were then grouped based on this classification.

The proportional relationship depended on the factor limiting MVF. When balance limited an MVF, the subsequent PAF was 80% of MVF. When strength limited an MVF, the subsequent PAF was 67%. Psychophysically acceptable forces were consistently related to MVF capacity and the biomechanical factor limiting that capacity. This finding provides a new insight to understand how psychophysical forces are selected from a biomechanical

perspective. Further, this proportional relationship may be exploited in future research to predict a PAF from knowledge of an MVF and the corresponding limiting factor.

6.2 - Introduction

The psychophysical approach has been used extensively for determining maximum acceptable exposure limits for work. Psychophysical limits have been reported for a variety of manual material handling tasks including: lifting, lowering, carrying, pushing, and pulling (Snook and Irvine, 1967; 1969; Snook, 1978; Ciriello and Snook, 1983; Ciriello et al., 1990; 1993; 1999; Snook and Ciriello, 1991; Ciriello, 2001); upper limb tasks including: wrist flexion/extensions, screw driving, hose insertions and wrist deviations (Snook et al., 1995; 1997; 1999; Ciriello et al., 2001; 2002; Moore and Wells, 2005; Andrews et al., 2008); and pinch, grasp, and finger pressing tasks (Nussbaum and Johnson, 2002; Potvin et al., 2006, McFall, 2008). The approach is attractive due to lower costs and the potential for faster application in industry than most biomechanical or physiological techniques (Ayoub and Dempsey, 1999). Also, psychophysical judgments are believed to take the whole job demand into consideration, integrating biomechanical and physiological factors (Karwowski and Ayoub, 1984). The major limitation of the approach is the assumption that subjectively selected workloads are below injury thresholds (Ayoub and Dempsey, 1999; Snook, 1999).

The lack of quantifiable support that psychophysically estimated forces relate to reducing injury risk is concerning. This concern has manifest due to a lack of research demonstrating a quantifiable link between psychophysical load selections and underlying mechanical loading (Thompson and Chaffin, 1993; Ayoub and Dempsey, 1999). In fact, psychophysical selection may not relate to known mechanical determinants of structural failure (Nicholson, 1989). Additional evidence has shown that workers choose loads despite their

generation of greater spinal compression forces than recommended biomechanical tolerance metrics (Chaffin and Page, 1994; Jorgensen et al. 1999).

However, recent evidence suggests that a quantifiable relationship may exist between psychophysical selections and biomechanical metrics. For example, during upper extremity exertions, participants subjectively chose workloads that did not result in electromyographic measures exceeding traditional markers indicative of increased injury risk (Moore, 1999; Cort et al., 2006; McFall, 2008). During lifting, Jorgensen et al. (1999) reported an association between the sagittal plane lumbar spine moment and the psychophysically acceptable lifting load. Further, joint loading (at the elbow, shoulder or trunk) remained below approximately 70% of the maximum possible joint moment when participants self selected their maximum psychophysically acceptable weight for a sustained static hold (Nussbaum and Lang, 2005).

Psychophysically determined maximum force also relates to maximum voluntary force (MVF). Using previously reported psychophysical data (Snook and Ciriello, 1991), and data collected in his lab, Potvin and colleagues have demonstrated that psychophysical limits for exertions completed once per minute are typically selected at approximately $2/3^{\text{rds}}$ of the MVF capability in a given posture (Potvin et al., 2006; Potvin, 2007; Andrews et al., 2008). Further, the decrease in psychophysically acceptable workloads that occurs as the duty cycle and cycle time is increased is predictable using a logarithmic curve fit anchored on the once per minute value of $2/3^{\text{rds}}$ of the MVF (Potvin, 2007). Though the simplicity of this general rule of $2/3^{\text{rds}}$ of the MVF for infrequent efforts would be attractive from a design or applied ergonomics perspective, further substantiation of its universality is needed.

Unlike psychophysical capacity, MVF capacity is guided by biomechanical parameters. The most influential parameters used in predicting MVF include whole-body balance and joint

strength (Gaughran and Dempster, 1956; Kroemer, 1974; Grieve 1979a; 1979b; 1983; Kerk et al., 1998; Seo et al., 2010; Fischer et al., submitted). During exertions where MVF is limited by balance, underlying joint demands are often low. Conversely, when strength limits MVF, the joint demand is maximal, specifically at the limiting joint (Fischer et al., submitted). Based on previous research demonstrating that psychophysically acceptable forces are at least in part related to joint demand (Jorgensen et al., 1999; Nussbaum and Lang, 2005), it is logical to assume that psychophysically acceptable forces will be related to the underlying limiting factor; as that factor dictates joint demand. Therefore, it is anticipated that psychophysically acceptable forces will be selected as a proportion of MVF (Potvin et al., 2006; Potvin, 2007; Andrews et al., 2008); however, that proportion will be specific to the underlying limiting factor. For example when joint strength is a limiting factor, joint demand is high and therefore psychophysically acceptable forces will be chosen as a smaller proportion to reduce the demand; this is in contrast to balance limited exertions, where joint demands are lower and the proportion can therefore be higher.

The purpose of this research was therefore to determine if estimates of psychophysical force were selected as a uniform proportion of MVF capacity independent of the exertion type or biomechanical factors limiting MVF capacity. The null hypothesis for this research states that psychophysically acceptable forces would not be selected at a different proportion of MVF dependent on the limiting factor.

6.3 – Methods

6.3.1 – Participants

Seventeen right-handed males were hired from a local temporary work agency to participate [mean age 41.4 ± 13.7 years; stature 1.74 ± 0.08 m; body mass 82.0 ± 14.7 kg].

Participants were required to have some general manufacturing experience (Appendix 2), be free of hand and forearm injuries within the past 6 months and have no sensitivity to ethanol on the skin. This study received ethical approval through the university board of ethics and all participants gave informed consent.

6.3.2 – Instrumentation

Force and postural information were collected to provide information to help address the research questions. Hand force was collected using an AMTI six degrees of freedom transducer (MC3A, AMTI MA, USA) rigidly fixed between a D-style handle and a clamp apparatus (Figure 6.1). The clamp allowed for the handle height to be adjusted according to the stature of each participant. Postural information was collected using an 8-camera VICON MX20 system (VICON, Oxford, UK). The motion of thirty-eight individual markers placed on anatomical landmarks was recorded, including the C7, suprasternal notch, xiphoid process, L5; bi-laterally, on the ear, acromion, lateral and medial epicondyle, radial and ulnar styloid, 2nd and 5th metacarpal, anterior superior iliac spine, greater trochanter, lateral and medial condyle, lateral and medial malleolus, heel, and 1st and 5th metatarsal. An additional eleven marker clusters fixated to rigid plates were attached over the sternum and bi-laterally on the top of the foot, shank, thigh, forearm, and upper arm to track segment motion during the experiment while reducing the effect of skin motion artifact. A static calibration frame established the relationship between the clusters and the calibration markers. The marker clusters were then used to track motions during the experimental trials. The marker cluster data and calibration frame were used to recreate virtual markers representing the landmarks above. These virtual markers were subsequently used to calculate joint centers and segment coordinate systems (Kingma et al., 1996). The joint center locations and segment coordinate systems were used as

inputs into the 3D static linked segment model described below. Specific details are also described in Chapter 4 and 5. Both force and motion were captured synchronously at 50 Hz.

6.3.3 – Experimental protocol

Participants completed exertions in three force directions (pull towards the body, down and medial) within three different postural conditions, for a total of nine different testing scenarios. For each scenario a handle, perpendicular to the grip, was positioned at shoulder height along the midline of the body, at a horizontal distance of 80% of the participants' upper limb length. These postures and force directions were chosen to 1) challenge different biomechanical limits (whole-body balance or joint strength) and 2) to simulate rubber window trim moulding tasks that have previously been shown to relate to injury development in the workplace (McClellan et al., 2009). The specific conditions (Figure 6.1) were:

1. **Shoulder Width Foot Placement (SWFP):** the participant stood with their feet shoulder width apart and their shoulders square to the force transducer and the right hand resting on the handle with a power grip. Their body was positioned at a distance equivalent to a comfortable arm's length (approximately 80 % of maximum reaching distance). Once the feet were comfortably in place, the researcher marked the foot locations on the floor in order to standardize foot position (Figure 6.1-A).
2. **Free Foot Placement (FFP):** similar to the condition above; however, the participant was free to place their feet in any orientation that was preferred (Figure 6.1-B).
3. **Upper Body Braced (UBB):** the participant stood as if they were going to complete a SWFP condition. A rigid frame was then slid in behind the participant until it was in contact with the posterior aspect of the participant. The frame was clamped into place using adjustable C-clamps and frame guides screwed into the floor. The participant was

then strapped to the board using two Velcro straps, one placed over the legs, just above the thigh clusters and a second over the torso just below the nipple line (Figure 6.1-C).

Experimental scenarios were presented to each participant in a random order. Prior to collecting the experimental data, two practice conditions duplicating the experimental conditions were completed to familiarize the participants with psychophysical estimation. The test conditions required participants to pull (practice trial 1) and press down (practice trial 2). Practice was provided as a means to help participants understand the self-selection of force approach and participants felt comfortable doing so without additional practice in the medial exertion direction.

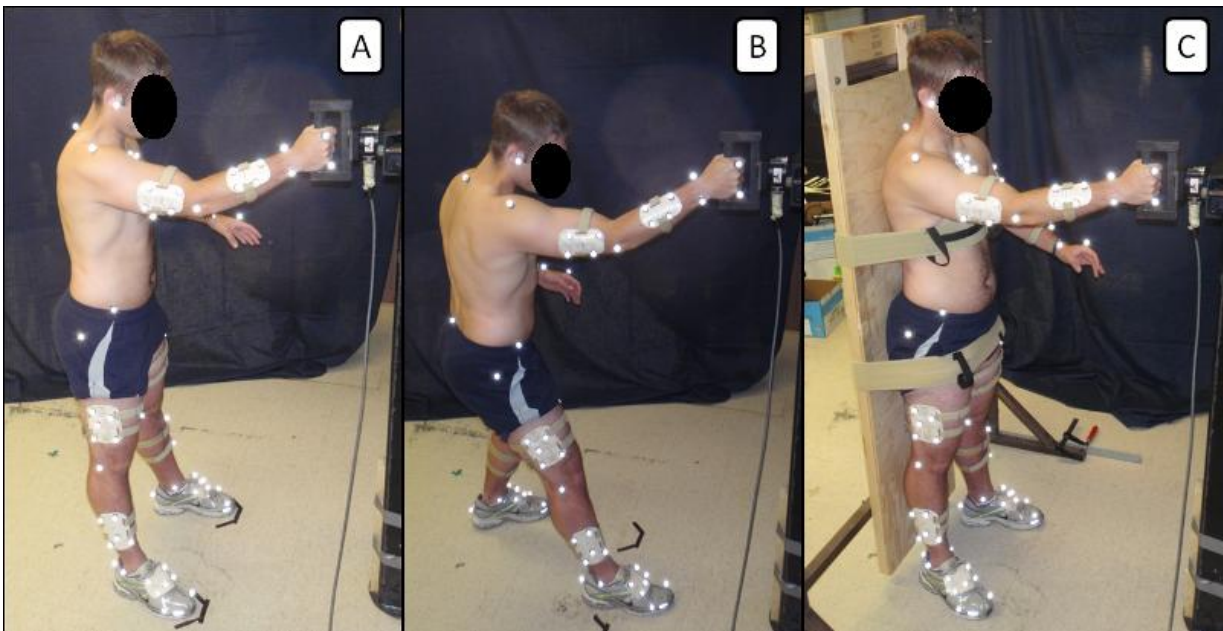


Figure 6.1 – The experimental conditions tested. Participants completed downward presses, pulls, and medial pushes in three conditions: 1) shoulder width foot placement (“SWFP” seen in frame A), free foot placement (“FFP” seen in frame B), and upper body braced (“UBB” seen in frame C).

A psychophysical load-adjust methodology was employed where participants were asked to exert force on the handle with magnitudes that they found acceptable (i.e. no signs of discomfort, numbness, or pain) at the predetermined cadence. The instructions were modified from Snook et al., (1995) (Appendix 1). Participants performed their maximal acceptable

exertions for 30 minutes per condition at one exertion per minute, sustaining each exertion for approximately one second. Two audible cues (spaced at one second apart) were presented every minute using a custom metronome (LabVIEW, National Instruments, Austin Texas, USA) to indicate when force was to be exerted.

Immediately before and following the performance of each condition, participants performed two 5-second maximum voluntary force (MVF) exertions in the requested force direction and condition. These MVF contractions provided an estimate of the maximum voluntary force in each condition and provided an indication of fatigue if the post MVF force was significantly less than the pre MVF force.

To ensure the participants were working within the guidelines provided in the instructions, discomfort ratings were taken 10 and 20 minutes into each scenario using a CR-10 scale (Borg, 1990). If participants responded with discomfort scores of 2 or more, the researcher re-read the instructions to them.

The total collection time required for each participant was approximately nine hours, divided over two days. The first day consisted of attaching the reflective markers to the participant, the completion of two practice conditions, followed by the completion of seven experimental scenarios. Participants were given scheduled lunch and rest breaks similar to a normal workday. The following day participants were again instrumented with reflective markers and completed the last two remaining experimental conditions. The remainder of day two was used to collect data for a complementary study.

6.3.4 – Data analysis

Data was analyzed to yield kinematic and kinetic information. All postural and force data were low-pass filtered using a dual pass Butterworth digital filter ($f_c = 4$ Hz, and $f_c = 15$

Hz respectively). Peak hand force in the desired direction was determined as the peak value resulting from a 500 millisecond moving window average passed over each minute of the filtered force trace. This process was used to identify the force magnitude during each of the 30 discrete exertions within each scenario. The 500 millisecond average helped to smooth the data to reduce the impact of jerking on the force cube when participants pushed, pulled or pressed. Figure 6.2 demonstrate how this process occurred using an representative data for one participant performing and pull in the FFP condition. The data points used to create the 500 millisecond average selected as the peak force were indexed in time allowing for the corresponding kinematics and six axis force data to be selected and averaged over the 500 millisecond window that aligned with the peak force data. The psychophysically acceptable force (PAF) was then calculated as the average force value recorded over the final five exertions within each test scenario. The resulting force was normalized to the highest peak force observed from the two pre- and post- scenario specific MVEs, yielding the normalized psychophysically acceptable force (%PAF).

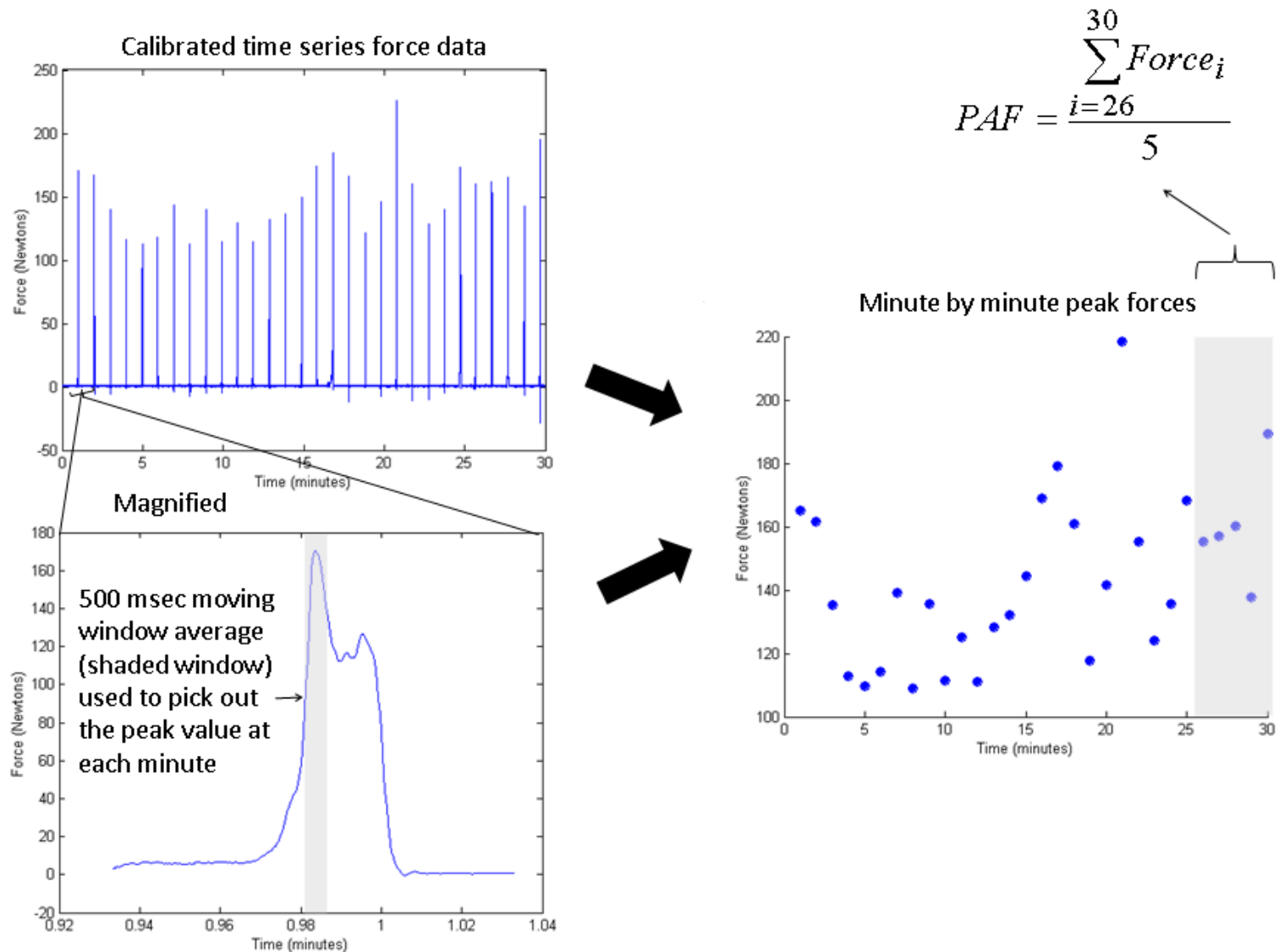


Figure 6.2 – The processing steps used to determine the psychophysically acceptable force using data from a representative participant performing a pull in the FFP condition.

In order to determine factors limiting force exertion (weakest link), the center of pressure (COP), elbow joint moments and shoulder joint moments were calculated and averaged across the four MVF trials for each condition. Shoulder moments were calculated by inputting the hand force and posture information into a 3D static linked segment model (adapted from Dickerson et al., 2007). The center of pressure was calculated using a top down approach by solving for the moment about the COP. To normalize COP between participants, the raw COP location was expressed as a percentage of the distance from the geometric centre of the base of support (BOS) to the edge of the base of support in both the anterior/posterior (A/P) and medial/lateral (M/L) directions (%COP). The BOS was defined using markers placed on the lateral malleolus, the tip of the 1st and 5th metatarsals, and at the posterior border of the calcaneus (Holbein-Jenny et al., 2007).

6.3.5 – Determining the weakest link

The weakest link was determined for each participant in each condition by comparing the %COP, elbow moment and shoulder moments calculated from the MVE trial that produced the highest hand force to the %COP and joint moment population thresholds described in the literature (Table 6.1). Balance thresholds were obtained from Holbein-Jenny et al. (2007). Flexion and extension limits for both the elbow and shoulder were calculated using the equations provided in Chaffin et al. (2006). Shoulder abduction, adduction, internal and external rotation strength limits were calculated using data from Schanne (1972) and Stobbe (1982). Based on the initial posture defined within the experimental conditions the following joint angles were used in estimating the joint strength limits: Elbow angle = 135°, Shoulder vertical angle = 90°, Shoulder horizontal angle and rotation angle = 0°. Though postures used to produce a maximum hand force deviated from the initial postures, the angles above deviated

by no more than 10° . The corresponding effect on the predicted population strength capacities would also vary based on the angle changes; however the effect on the calculated population strength was less than 10 Nm. Wrist strength was not considered as a limiting factor based on previous research demonstrating that wrist strength is typically only potentially a limiter when it is in a non-neutral position (Al-Eisawi et al., 1998). In this study participants maintained a neutral wrist posture while exertion force on the handle.

Once the limits were established, MVF exertions were then classified as limiting if the experimentally determined value (COP or joint specific moment) fell within one standard deviation of the population limit threshold, or exceeded the limit. If more than one weak link was identified during this process the limit that exceeded the threshold by the greatest percentage was selected as the primary limiting factor. When no limiter could be found, it was assumed to be trunk strength in all UBB conditions (where balance could not be limiting by design) or undefined for SWFP and FFP conditions (where friction or trunk strength may have been limiting). Following the weakest link classification process, each %PAF value was sorted into one of three groups – strength, balance, or undefined based on the weakest link exposed during the corresponding MVF trials.

Table 6.1 – Threshold limit values used to determine if balance or joint strength was limiting an MVF exertion.

Threshold limits		
Variable	Population Limit	Standard Deviation
Elbow Flexion (Nm) ¹	64.33	15.81
Elbow Extension (Nm) ¹	56.16	11.30
Shoulder Flexion (Nm) ¹	64.68	14.95
Shoulder Extension (Nm) ¹	101.40	20.41
Shoulder Internal Rotation (Nm) ²	39.12	11.25
Shoulder External Rotation (Nm) ²	34.88	7.91
Shoulder Adduction (Nm) ²	83.83	24.53
Shoulder Abduction (Nm) ²	100.26	27.32
Frontwards COP Excursion Limit (%BOS) ³	0.58	0.17
Backwards COP Excursion Limit (%BOS) ³	0.55	0.13
Right Side COP Excursion Limit (%BOS) ³	0.66	0.13
Left Side COP Excursion Limit (%BOS) ³	0.67	0.13
¹ Chaffin et al., 2006		
² Schanne 1972; Stobbe 1982		
³ Holbein-Jenny et al., 2007		

6.3.6 – Statistical analysis

To test the null hypothesis that %PAF is uniform between exertions limited by balance and exertions limited by joint strength, an independent samples Wilcoxon rank sum test was used. A non-parametric test was selected to avoid the assumption of normality required with a parametric counterpart. The test was selected to detect any significant differences in the dependent variable %PAF between the two limiting factors (balance and strength). To determine if the psychophysical protocol resulted in fatigue, a repeated measures ANOVA was used to detect significant differences in the MVF force between the pre and post trials for each of the nine experimental conditions. A Greenhouse-Geisser correction was used to protect against violations of the sphericity assumption. Significance levels were set at $p < 0.05$ for all

tests. All statistical processing was completed using SPSS software (SPSS INC., Chicago, IL, USA).

6.4 – Results

Of the 129 exertions classified, 33 (26%) were limited by balance, 76 (59%) were limited by joint strength and 20 (15%) were undefined. The Wilcoxon test identified that the %PAF were significantly different, $z = -4.27$, $p < 0.001$. When balance was limiting %PAF values were $83 \pm 19\%$. When joint strength was limiting %PAF values were $67 \pm 17\%$ (Figure 6.3). The repeated measures ANOVA did not detect a significant decrease in hand force between pre and post conditions indicating no loss in MVF capacity as a result of the psychophysical protocol (Figure 6.4).

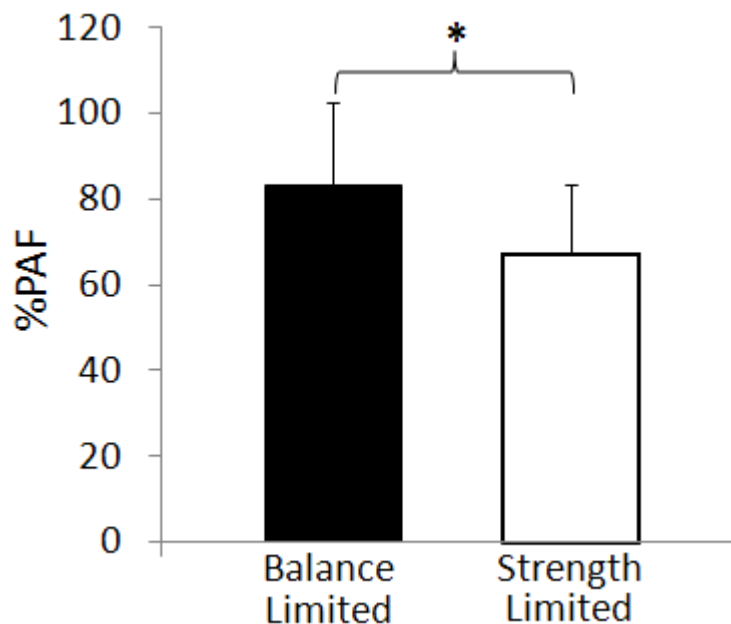


Figure 6.3 – The psychophysically acceptable force as a percentage of the maximum voluntary force (%PAF) during exertions limited by balance and those limited by strength. The significant difference is denoted by an asterisk ‘*’ ($z = -4.27$, $p < 0.001$).

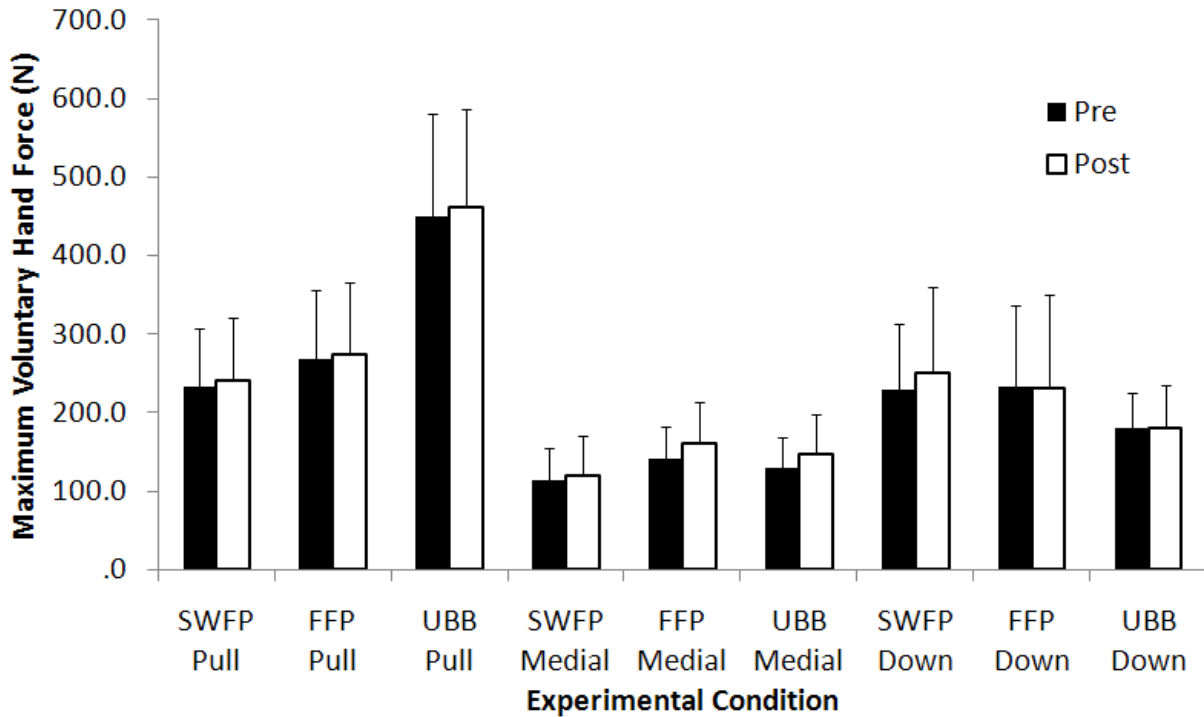


Figure 6.4 – The maximum voluntary force measured before (pre) and after (post) exposure to a thirty minute psychophysical estimation protocol in each of the nine experimentally defined conditions. No significant differences were found between the pre and post force levels within each experimental condition. SWFP – shoulder width foot placement, FFP – free foot placement, UBB – upper body braced.

The limiting factors were assigned by comparing the %COP and joint moments from individual trials to population thresholds. Table 6.2 documents the number of times an exertion was limited by a specific factor within each of the experimental conditions. Though each condition (9) should have 17 exertions classified (n=17), this was not possible. In 24 of the 153 total exertions, motion data was not available. Throughout the work day, the rigid plates occasionally shifted over the course of a thirty minute trial. Although rigid plates were checked often, and several static calibrations were performed, in 24 instances the rigid plates shifted such that the corresponding estimated joint centers and segment coordinate systems were not indicative of actual postures. This is discussed in more depth in the limitations section.

Table 6.2 – Classification of maximum voluntary exertions based on the limiting factor stratified by experimental condition. When no limiter was uncovered it was classified as undefined. All UBB exertions not limited by the upper limb were assumed to be limited by trunk strength as balance and friction limitations were removed as possible constraints through the use of bracing. The number of exertions within each condition (“n”) differs as a result of missing data.

Direction	Condition	n	Balance				Strength									Undefined
			Front	Back	Left	Right	Elb Flex	Elb Ext	Sho Flex	Sho Ext	Sho Ab	Sho Add	Sho IR	Sho ER	Trunk	
Down	FFP	14		3						6		2	1		1	1
Down	SWFP	15		3						5		2	4			1
Down	UBB	15						1		4		7			3	
Medial	FFP	13				2				1		3				7
Medial	SWFP	14					1		2			3				8
Medial	UBB	15						1				4			10	
Pull	FFP	12	11													1
Pull	SWFP	16	14													2
Pull	UBB	15					4			2		3			6	
Elb - Elbow, Sho - Shoulder																
Flex - Flexion , Ext - Extension, Ab - Abduction, Add - Adduction, IR - Internal Rotation, ER - External Rotation																

6.5 – Discussion

These findings reject the null hypothesis and show that the proportion of MVF exerted in the psychophysical load-adjust protocol is different between situations when balance limits force and where joint strength limits force. This is an important finding for two reasons. First, it provides additional quantitative evidence to support a relationship between psychophysical estimates and biomechanical factors (Nussbaum and Lang, 2005; Potvin, 2007). Second, this finding encourages the possibility of using a weakest link classification approach to help facilitate predictions of psychophysically acceptable forces from known MVF estimates (Potvin, 2007).

This research used a novel approach for classifying psychophysical chosen exertions based on the biomechanical factor limiting a corresponding MVF. During un-braced pulling, 25 of 28 MVF exertions were limited by balance, while 21 of 29 downward exertions were limited by joint strength. Previous research classifying limiters for MVF exertions have found similar results, in that pushing and pulling were balance limited (33 of 34 exertions) while lifting and downward pressing were limited by joint strength (22 of 30 exertions), (Kerk et al., 1998). Extending these findings, if we assume horizontal exertions along the sagittal plane at or near shoulder height are likely balance limited, and vertically aligned exertions close to the body are likely strength limited, we can classify previously documented psychophysical exertions as being balance or strength limited based on the exertion direction and task condition. Table 6.3 presents data from Snook and Ciriello's classic 1991 paper expressed as the psychophysically acceptable force at one exertion per minute, the corresponding MVF value assumed from the acceptable force value at the once per day rate and the resulting %PAF for males.

Table 6.3 – The %PAF calculated as a ratio between psychophysically acceptable forces at one exertion per minute and at one exertion per day (assumed to represent the MVF) (Snook and Ciriello, 1991), classified based on the theoretical limiting factor.

Balance Limited				Joint Strength Limited			
Pull at 144 cm - 2.1 m pull (kg)				Lower - Knuckle height to floor level - width 34 cm - distance 25 cm (kg)			
%ile	1 / min	1 / day	%PAF	%ile	1 / min	1 / day	%PAF
90	18	23	78%	90	20	32	63%
75	22	28	79%	75	28	46	61%
50	26	33	79%	50	37	60	62%
25	31	39	79%	25	46	75	61%
10	34	44	77%	10	54	89	61%
Push at 144 cm - 2.1 m pull (kg)				Lift- Floor level to knuckle height - width 34 cm - distance 25 cm (kg)			
%ile	1 / min	1 / day	%PAF	%ile	1 / min	1 / day	%PAF
90	25	31	81%	90	18	27	67%
75	32	41	78%	75	26	38	68%
50	40	51	78%	50	35	52	67%
25	47	61	77%	25	44	65	68%
10	55	70	79%	10	52	76	68%
Mean % PAF for Balance Limited			79%	Mean % PAF for Joint Strength Limited			65%
%ile - percentage of industrial population accommodated							
1 / min - Psychophysically acceptable force at one exertion per minute							
1 / day - Maximum Voluntary Force							
% PAF - Psychophysically acceptable force / Maximum voluntary force X 100%							

The %PAF relationships calculated from the Snook and Ciriello (1991) data support the current research proposing that psychophysically acceptable force selection is related to the MVF and dependent on the biomechanical factor most likely limiting the MVF. Further evidence to support this relationship is discernable from comparing psychophysically acceptable initial pulling forces of 425 N (Ciriello et al., 1990) to the population maximum voluntary pulling force capacity of 537 N (Mital and Kumar, 1998), yielding a %PAF of 79%, where both measures were taken at a height of approximately 1 metre.

Other past psychophysical research also demonstrates this proportional relationship when viewed using a weakest link, or limiting factor approach. Work by Andrews et al., (2008) measured psychophysically acceptable forces for a variety of hose insertion tasks (varied by task location and the direction of the hose insertion). They calculated the %PAF in each condition and found a consistent %PAF value of 63%. They did not determine what the potential limiting factor might be, though it is likely that joint strength in the upper limb is a dominant weak link in these exertions. The orientation of the exertions were such that elbows were often at 90° of flexion and the shoulder was often at 0° up to a maximum of 45° of elevation from the vertical. From a biomechanical perspective these positions would: direct pushing and pulling forces through the whole body center of mass – protecting against a balance limiter; while increasing the shoulder and elbow moments, increasing the probability of a strength limitation.

Other researchers have reported %PAF values below the 67% thresholds reported in this study. Nussbaum and Johnson (2002) reported %PAF values of 45.6 ± 11.3 % and 35.4 ± 15.1 % for finger pressing tasks using the index finger and thumb respectively. Similarly, Potvin et al. (2006) reported %PAF values of 48.1 ± 16.1 %, 61.8 ± 15.7 %, and 63.4 ± 16.5 % during oblique grasping, finger pressing and pulp pinching respectively. Though these values are less than those reported here (83 ± 19 % and 67 ± 17 % for balance and strength limited exertions) the variability is consistent between studies. A weakest link explanation may help provide context for these differences. For example, both previous studies examined finger pressing and reported considerably different %PAF values. This may be a result of posture. In the study by Nussbaum and Johnson (2002) participants extended their index finger, with the hand pronated and the palm facing down, and then proceed to press down in a vertical

direction, perpendicular to the long axis of the extended index finger. Biomechanically, this causes a moment about each of the joints along the index finger, and so on traveling proximally along the limb. Therefore finger joint strength could be a limiting factor if the moment reached a finger joint limit threshold. Conversely, Potvin used a similar posture; but asked participants to push horizontally, along the axis of the finger. Biomechanically, the force line of action now passes through the joints of the index finger, greatly reducing any moments at those joints. The limiting joint may then be further along the kinematic chain, possibly the elbow or shoulder. It is likely that each joint may have a different resolution, whereby a PAF of 67% is not uniform across all joints, but each joint may have a more specific threshold. These data do not provide the detail necessary to answer that question, but future work may help address this issue in more detail.

The proportional relationship between psychophysically acceptable forces and MVF in exertions limited by strength is consistent with previous findings. Many theories of injury causation describe how overexertion may lead to injury (McGill, 1997; Kumar, 2001). In addition, participants can perceive shoulder and low back muscular effort, measured in terms of joint strength (Nussbaum and Lang, 2005; Dickerson et al., 2006; Nastasia et al., 2007). Therefore a psychophysically acceptable exertion force may be selected in part to provide a margin of safety (Kumar and Mital, 1992) to protect against joint strength overexertion. However, as the exposure duration, cycle time or frequency of a task changes, this relationship is also likely affected and the perception of joint strength alone may not be enough to protect against other modes of injury (Kumar and Mital, 1992). It remains unclear why a margin of 67% of the MVF capacity is chosen specifically; however, evidence supports that psychophysical exertion levels at one exertion per minute, in strength limited conditions, are

likely chosen to provide a margin of safety to protect against joint strength related overexertion.

The proportional relationship between psychophysical forces and MVF in exertions limited by balance is more challenging to explain. Participants do not extend their COP all the way to the BOS edge during loaded and unloaded leaning (Holbein and Chaffin, 1997; Holbein and Redfern, 1997; Lee and Lee, 2003; Holbein-Jenny et al., 2007); rather they limit COP excursions to a smaller region of the base of support termed the functional base of support (FBOS). Hence, when balance is most challenged, a balance safety margin already exists. Therefore, during psychophysical exertions in conditions limited by balance it is unclear why an additional margin of safety is introduced. It may be that the additional safety margin is related to the muscular demand required to maintain balance near the edges of the FBOS. Indeed, recent research has shown that sustained isometric pushing efforts ceased as a result of postural muscle fatigue and not fatigue in the prime movers (Le Bozec et al., 2004). In the current study, muscle activity or joint loading were not determined for the lower limbs, prohibiting a more detailed understanding of why the 80% proportionality persists for psychophysical exertions in conditions where the MVF capacity is limited by balance.

This study had inherent limitations. The major assumption is that participants adopt the same posture and exert forces in the same directions during the psychophysical exertions as they do during the MVF exertions, and are thereby being limited by the same factor. Although this was not quantitatively assessed, qualitatively participants remained in a similar posture during both MVF and psychophysical exertions. Errors arising from the postural data collection, such as inter-trial skin motion and marker placement accuracy, and the use of population-based anthropometric tables to estimate segment masses, cause some uncertainty in

the COP calculation and the linked segment modeling outcomes. These effects were reduced through the use of rigid clusters to help reduce potential artifacts in the motion capture data (Kingma et al., 1996) and the error from the use of anthropometric tables should be randomly distributed across participants. To further reduce this effect, motion data was discarded for 24 of 153 trials (16%). These motion data were discarded as the rigid clusters shifted during collection (due to sweat, by being bumped, or due to the straps in the UBB conditions) affecting the re-created anatomical landmarks. The trunk cluster was most often affected. It was assumed that the trunk was a rigid segment, and therefore one marker cluster was used to generate virtual landmarks required to describe the trunk posture. This assumption did not always hold as some participants flexed through the spine and in some cases bumped the cluster, affecting the recreation of the virtual landmarks and subsequent trunk postural description. This was determined qualitatively by visually assessing all re-created anatomical motion data. Only males participated in this study as a result of financial constraints. The research budget did not permit hiring more than seventeen workers and a decision was made to have a larger sample size for a homogeneous population, rather than smaller sample sizes in diverse populations.

The approach used to classify exertions a balance or strength limited is based on population threshold data. This is a limitation of this work as it is difficult to interpret an individual response (COP or joint strength) in the context of population distributions. To help account for this the thresholds were established as one standard deviation below the population mean. This approach sets a conservative threshold, which may have caused specific joints to be selected as limiting although that individual may have had capacity well beyond that threshold. However, this would have more impact on selecting the appropriate joint where

strength is limiting rather than impacting the classification between strength or balance limited, affecting the overall findings of the study.

6.6 – Conclusions

Psychophysically determined acceptable forces for unilateral exertions at once per minute were selected as a proportion of the MVF capacity for that exertion. By classifying MVF exertions as being balance or strength limited, the corresponding %PAF were sorted and found to be significantly different. Psychophysical loads were selected at approximately $2/3^{\text{rds}}$ of the MVF capacity during strength limited exertions, while they were selected at $4/5^{\text{ths}}$ when the MVF capacity was limited by balance. These results demonstrate that a biomechanical weakest link classification strategy may help to elucidate how workers select psychophysically acceptable forces dependent on their maximum capacity. This relationship may help direct future research efforts towards: 1) predicting psychophysical limits, and 2) understanding the relationship between psychophysical thresholds and biomechanical links to tissue damage or injury.

Chapter 7 – Predicting Psychophysically acceptable hand forces by scaling predictions of maximum hand force capacity based on the biomechanically limiting factor

Steven L Fischer, Clark R Dickerson and Richard P Wells

7.1 – Overview

This paper describes a novel approach to predict psychophysically acceptable forces during unilateral exertions. Traditionally, psychophysically acceptable forces (PAFs) have been determined experimentally or predicted using task-specific tables of regression equations. In this study a biomechanical model is used to first estimate the maximum voluntary force (MVF) capacity based on a weakest link approach, and then derives a predicted PAF by multiplying the MVF by a scaling factor that is specific to the biomechanical factor that is limiting the MVF capacity: whole body balance or joint strength. This study provides an evaluation of this approach using one-handed horizontal pulling and downward pressing tasks, performed at a frequency of once per minute.

Experimental hand force data showed that participants chose PAFs at $2/3^{\text{rds}}$ of their MVF capability when they were limited by joint strength and chose $4/5^{\text{ths}}$ of the MVF capacity when they were limited by balance. This finding supported the use of a limiting factor scaling approach to predict PAFs from MVFs. When using these proportions to predict PAFs from model estimated MVFs, the model underestimated PAFs by 24% and 43% during downward and pulling exertions respectively. This underestimation is the result of the biomechanical model underestimating MVF capacity by 23% and 34% prior to scaling MVFs to predict PAFs. This model provides an innovative approach for predicting PAFs.

7.2 – Introduction

The psychophysical approach has been used extensively to determine maximum acceptable exposure limits for work. Psychophysically acceptable limits have been reported for a variety of manual material handling tasks including: lifting, lowering, carrying, pushing, and pulling (Snook and Irvine, 1967; 1969; Snook, 1978; Ciriello and Snook, 1983; Ciriello et al., 1990; 1993; 1999; Snook and Ciriello, 1991; Ciriello, 2001); upper limb tasks including: wrist flexion/extensions, screw driving, hose insertions and wrist deviations (Snook et al., 1995; 1997; 1999; Ciriello et al., 2001; 2002; Moore and Wells, 2005; Andrews et al., 2008); and pinch, grasp, and finger pressing tasks (Nussbaum and Johnson, 2002; Potvin et al., 2006, McFall, 2008). Psychophysically derived exposure threshold limits are appealing because they are more time and cost effective to apply in industry than many biomechanical or physiological techniques (Ayoub and Dempsey, 1999). Conversely these thresholds are limited in that there is little support that working below psychophysical thresholds will help reduce injury risk (Ayoub and Dempsey, 1999; Snook, 1999).

Improving psychophysical thresholds as an option to reduce injury risks in the workplace is critical to the advancement of the approach. It requires a more detailed understanding of the relationships between psychophysical capacity, corresponding job demand and the incidence and severity of musculoskeletal disorders (MSDs). In an ideal situation, an epidemiological paradigm would be well situated to investigate these relationships (Ayoub and Dempsey, 1999) given data on the psychophysical capacity, job demand and MSD severity. The underlying challenge is in readily determining psychophysical capacities for a range of work scenarios. Though psychophysical capacity estimates are readily obtainable in a lab environment, they are less easy to obtain in the field. Rather,

psychophysical capacity predictive models (PCPM) may provide a means to forecast capacities for a number of workers in a number of jobs much more efficiently, facilitating the epidemiological research required to support or refute the future use of psychophysically based exposure thresholds (Ayoub and Dempsey, 1999).

Several approaches have been used to try to develop psychophysical capacity prediction models (PCPM). Techniques have included stepwise regression (Genaidy et al., 1988), prediction using Steven's Power Law (Nussbaum and Johnson, 2002), and the use of fuzzy logic (Karwowski and Ayoub, 1984). Regression has been used most frequently (Ayoub et al., 1980; Genaidy et al., 1988). Regression equations provide a straightforward way to predict capacity, but the equations are task specific and data set dependent. The power law approach adopted by Nussbaum and Johnson (2002) is also straightforward to use. However, the approach does not capture any individual parameters, assuming that the capacity is the same for everyone and based only on the force required and the frequency of force application. The fuzzy logic approach is novel to psychophysical capacity modeling, but remains challenging to integrate into a useful field technique (Genaidy et al., 1988).

Predictive approaches based on biomechanical principles rather than correlations and associations may be more effective. The historical dominance of correlation and association based PCPMs were likely due to the lack of evidence for any alternative underlying principles. However, recent work has reported a general rule that psychophysical forces for infrequent (one exertion per minute) exertions are selected at approximately $2/3^{\text{rds}}$ of the maximum voluntary force (Potvin et al., 2006; Potvin, 2007; Andrews et al., 2008; Chapter 6). Further, this general rule has been extended to demonstrate that the proportionality is dependent on the mechanical factor most likely limiting MVF capacity (Chapter 6). When an MVF exertion is

likely limited by joint strength, workers will subsequently choose to psychophysically work at about $2/3^{\text{rds}}$ of their MVF capacity in that task. However, when the MVF exertion is limited by whole body balance, workers will alternatively choose to psychophysically work at about $4/5^{\text{ths}}$ of the MVE capacity in that task. This proportional relationship allows for psychophysical limits to be predicted given the MVF and the factor most likely limiting the MVF.

A model that predicts MVF capacity and the biomechanical weakest link, was recently developed (Chapter 4). The model was adapted to include these psychophysical proportions to additionally yield predictions of the psychophysically acceptable capacity. The purpose of the study discussed in this chapter was to evaluate if the adapted model predicts psychophysical capacities that approximate empirical psychophysically derived acceptable forces for unilateral pulling and downward pressing tasks. Further the experimental data were used to determine if the $2/3^{\text{rds}}$ and $4/5^{\text{ths}}$ proportionalities persisted.

7.3 – Methods

7.3.1 – Participants

Seventeen right-handed males were hired from a local temporary work agency to participate in this study [mean age 41.4 ± 13.7 years; stature 1.74 ± 0.08 m; body mass 82.0 ± 14.7 kg]. Participants were required to have some general manufacturing experience (Appendix 2), be free of hand and forearm injuries within the past 6 months and have no sensitivity to ethanol on the skin. This study received ethical approval through the university board of ethics and all participants signed an informed consent.

7.3.2 – Instrumentation

Force and posture information were collected for this study. Force was collected using an AMTI six degrees of freedom transducer (MC3A, AMTI MA, USA) rigidly fixed between a D-style handle and a clamp apparatus (Figure 7.1). The clamp allowed for the handle height to be adjusted according to the stature of each participant. Kinematic information was collected using an 8-camera VICON MX20 system (VICON, Oxford, UK). Motion was recorded from thirty-eight individual markers placed on anatomical landmarks, including the C7, suprasternal notch, xiphoid process, L5; bi-laterally, on the ear, acromion, lateral and medial epicondyle, radial and ulnar styloid, 2nd and 5th metacarpal, anterior superior iliac spine, greater trochanter, lateral and medial condyle, lateral and medial malleolus, heel, and 1st and 5th metatarsal. An additional eleven marker clusters fixated to rigid plates were attached over the sternum; bi-laterally, on the top of the foot, shank, thigh, forearm, and upper arm. Marker clusters were used to track segment motion during the experiment to reduce the effect of skin motion artifact. A static calibration frame established the relationship between the clusters and the calibration markers over the anatomical landmarks, and subsequently joint centers and segment coordinate systems were described (Kingma et al., 1996). These data were used to describe the whole body posture required to drive the model. Both force and motion were captured synchronously at 50 Hz.

7.3.3 – Experimental protocol

Participants performed a repetitive horizontal pulling task and a downward pressing task (Figure 7.1) each for thirty minutes at one repetition per minute. For each scenario the handle was positioned at chest height along the midline of the body. An obstruction was placed in front of the participant. The distance from the obstruction to the handle on the force cube

was equivalent to 75% of the participant's upper limb length. The obstruction was used to force the participant to reach to the handle to simulate reaching over a surface in a working environment. They were instructed not to make contact with the hip bar or with the foot plate. Participants were then able to self select their desired working postures within the confines of those instructions.

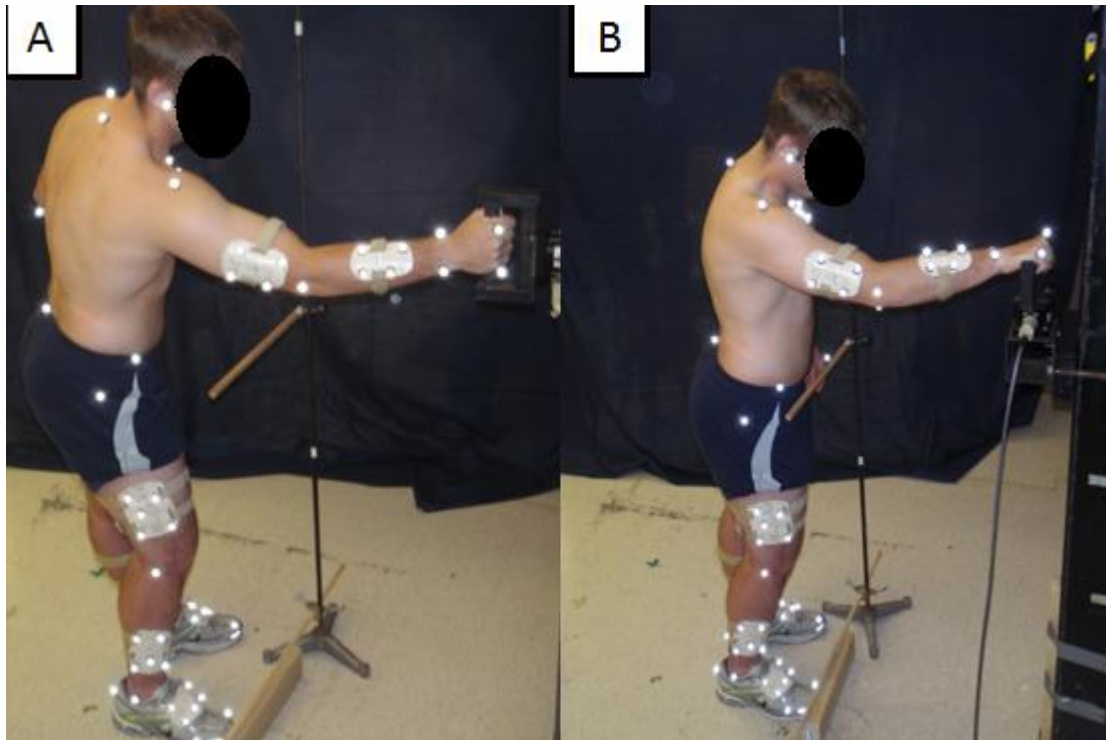


Figure 7.1 – The experimental conditions tested. Participants completed horizontal pulls ('A') and downward presses ('B'), at one exertion per minute for thirty minutes, at their psychophysically acceptable force level. Participants also completed maximum voluntary exertions in these positions, both prior to- and following the psychophysical exertions. The obstruction was used to oblige participants to reach for the handle simulating a work environment.

A psychophysical methodology was employed where participants were asked to exert force on the handle with magnitudes that they found acceptable (i.e. no signs of discomfort, numbness, or pain). The instructions were modified from Snook et al., (1995) (Appendix 1). Participants completed the psychophysically acceptable exertions for 30 minutes per condition at a cycle time of one exertion per minute, sustaining each exertion for approximately one

second. Two audible cues (spaced at one second apart) were presented every minute using a custom metronome software program (LabVIEW, National Instruments, Austin Texas, USA) to indicate when force was to be exerted.

To ensure the participants were working within the guidelines provided in the instructions, discomfort ratings were taken at 10 and 20 minutes into each scenario using a CR-10 scale (Borg, 1990). If participants responded with discomfort scores of 2 or more, the researcher re-read the instructions to them.

Before and following the performance of the psychophysical force estimation protocol, participants performed two 5-second maximum voluntary force (MVF) exertions in the requested force direction and condition. These MVF contractions provided a measure of the maximum capable force in each scenario for each participant.

Experimental scenarios were presented to each participant in a random order. This study took place as part of a larger research study, where participants had completed psychophysical exertions in the lab for a full eight-hour day prior to completing the exertions described here. They were familiar with the psychophysical paradigm and comfortable in the lab environment.

7.3.4 – Experimental data analysis

Data was analyzed to yield kinematic and kinetic information. All postural and force data were low-pass filtered using a dual pass Butterworth digital filter ($f_c = 4$ Hz, and $f_c = 15$ Hz respectively). Peak hand force in the desired direction was determined as the peak value resulting from a 500 millisecond moving window average passed over each minute of the filtered force trace (Appendix 3A). This process was used to identify the force magnitude during each of the 30 discrete exertions within each scenario. The 500 millisecond average

helped to smooth the data to reduce the impact of jerking on the force cube when participants pushed, pulled or pressed. The psychophysically acceptable force (PAF) was then calculated as the average force value recorded over the final five exertions within each test scenario. The resulting force was normalized to the highest peak force observed from the two pre- and post-scenario specific MVEs, yielding the normalized psychophysically acceptable force (%PAF).

The data points used to create the thirtieth and final 500 millisecond average, within each condition, were indexed in time allowing for the corresponding posture data to be selected and averaged over the same range, aligning with the peak force data. These data served to drive the force prediction model.

7.3.5 – Model description

A hand force capacity prediction model, previously described, was used to generate predictions of the MVF and limiting factor (Chapter 4). The original model computed MVF capacity and predicted the weakest link by using the inputs of posture, stature, body mass and prescribed exertion direction. Using these inputs, joint strengths (moments) and whole body balance requirements (center of pressure relative to the base of support) are incrementally calculated as the assumed hand forces increased from 1 to 1000 N. The resultant joint strength and center of pressure (COP) locations are then compared to stochastically generated constraint thresholds for joint strength and COP excursion limits. This process is used to identify the maximum hand force achievable without exceeding a constraint. Using a Monte Carlo simulation the constraint generation and threshold comparisons were repeated for 1000 iterations. At each iteration the constraint thresholds are selected from a normal distribution of possible constraint values based on experimental data previously reported in the literature (Schanne, 1972; Stobbe, 1982; Holbein and Chaffin, 1997). Following the simulation, the

model output a predicted MVF and standard deviation, while also describing the constraint that most frequently limited MVF capacity through the simulation process.

The psychophysical predictions were computed within the Monte Carlo simulation processes, where the maximum hand force determined at each iteration was multiplied by 0.67 (2/3^{rds}) if a strength based constraint limited capacity or by 0.8 (4/5^{ths}) if a balance based constraint limited capacity. Following the Monte Carlo simulation the updated model also output the PAF and the corresponding standard deviation.

7.3.6 – Model evaluation

Experimentally obtained posture information was used as an input into the model, along with the participant's specific stature and body mass and the direction of the exertion, extracted from the force data. Participants were instructed to exert forces in a given direction (i.e. horizontal pull); however participants often exert additional off-axis forces (de Looze et al., 2000; Granta and Bennett, 2005; Hoffman et al., 2007). This change in force direction has been shown to affect the ability to predict MVF capacity (Chapter 5). The actual force direction vector was used as an input in this case to parse out the effect that misrepresenting the desired force vector may have in estimating both MVF and psychophysically acceptable forces (Chapter 5).

Model evaluation was performed in three ways: 1) Evaluation of the proportional relationships: the measured PAFs were categorized based on the model predicted limiting factors then divided by the corresponding measured MVF to test if %PAF values were indeed selected a different magnitudes dependent on the weakest link. This comparison was evaluated statistically using a Mann Whitney U test. A non-parametric test was selected to avoid the assumption of normality required with a parametric counterpart, and the Mann

Whitney U was selected instead of the Wilcoxon sum of ranks tests because each group had less than 20 observations.; 2) Maximum magnitude similarity: the model estimates of MVF capability were compared to the pre and post measured MVF values obtained during the experiment to demonstrate the ability to predict MVF capacity; 3) Psychophysical magnitude similarity: measured PAFs were compared to predicted PAFs in each condition to evaluate if predictions under- or overestimated experimentally determined values. The magnitude based measured versus predicted comparisons (evaluations 2 and 3) were made using two separate repeated measures ANOVA's. A Greenhouse-Geisser correction was used to protect against violations of the sphericity assumption. Significance levels were set at $p < 0.05$ for all tests. All statistical processing was completed using SPSS software (SPSS INC., Chicago, IL, USA).

7.4 – Results

The proof of principle evaluation supported the use of the 2/3^{rds} and 4/5^{ths} proportional relationships (Figure 7.2). In exertions classified as balance limited, participants chose %PAF values at $79.8 \pm 20.9\%$, significantly higher ($U=30$, $U_{crit} = 37$, $n_1=9$, $n_2=16$) than those limited by strength, chosen at $64.8 \pm 21.6\%$. The magnitude similarity evaluations demonstrated that the model predictions underestimated both experimental MVFs and PAFs (Figure 7.2). The significantly model significantly underestimated psychophysical forces by 24% during downward exertions ($F=6.649$, $p=0.026$, partial $\eta^2 = 0.377$) and by 43% during pulling exertions ($F=6.630$, $p=0.026$, partial $\eta^2 = 0.376$). However, the underestimations of PAF values were primarily a result of the underestimation of MVFs (23% for downward exertions ($F=12.315$, $p=0.001$, partial $\eta^2 = 0.26$) and 34% for pulling exertions ($F=36.218$, $p < 0.001$, partial $\eta^2 = 0.509$)).

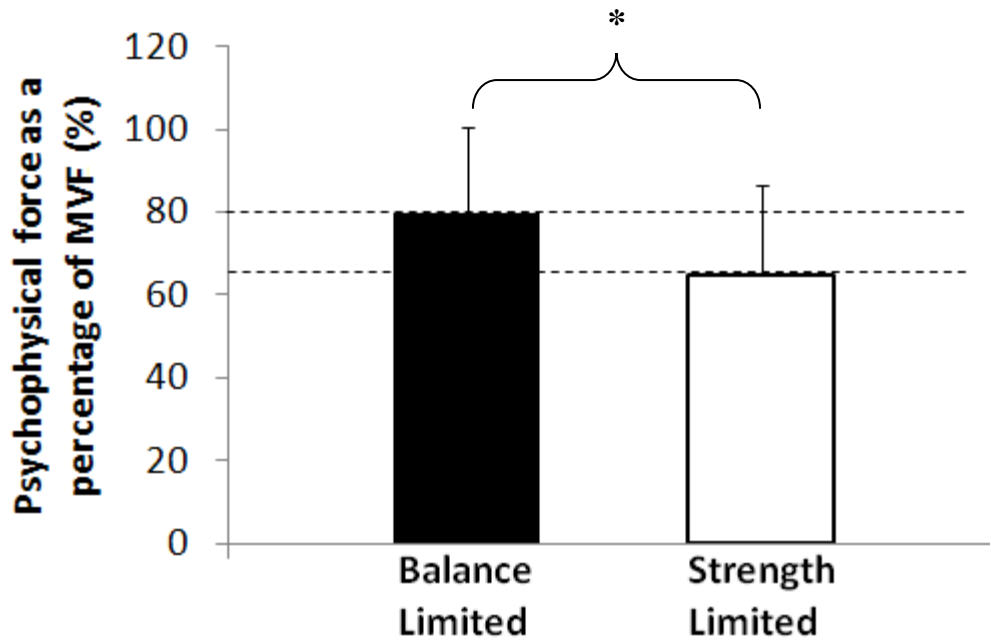


Figure 7.2 – The psychophysical acceptable force (mean and standard deviation) as a percentage of the maximum voluntary force (MVF) during exertions limited by balance and those limited by strength. Chapter 6 suggests that strength limited exertions should be selected at approximately $2/3^{\text{rds}}$ of MVF, where balance limited should be selected at $4/5^{\text{ths}}$ of the MVF. The significant difference is denoted by an asterisk ‘*’.

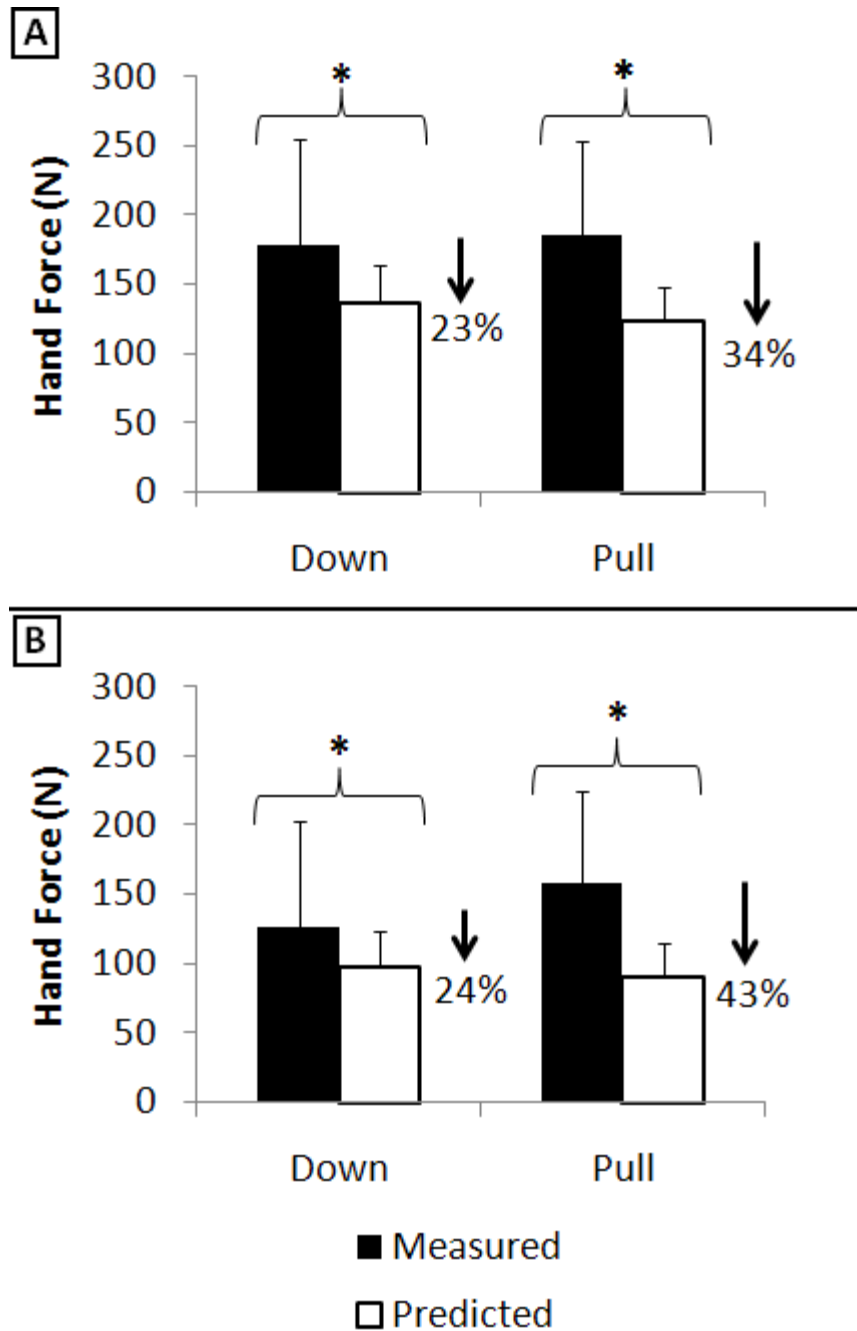


Figure 7.3 – A comparison between measured and predicted hand forces (means and standard deviations). Comparisons made during maximum voluntary force exertions are shown in ‘A’, and comparisons made during psychophysically acceptable exertions are shown in ‘B’. Significant differences are denoted with an asterisk ‘*’. Arrows and percentages describe the relative difference between the measured and predicted values.

7.5 – Discussion

The current method for predicting psychophysically acceptable forces provides a robust approach based on an underlying testable principle. The underlying principle poses that PAFs at one exertion per minute are proportional to the MVF capability, where the proportionality is dependent on the biomechanical weakest link governing MVF capability. The proof of principle evaluation provides additional support to justify the use of this relationship predicting PAFs (Chapter 6). The magnitude similarity comparisons demonstrate that the current approach conservatively estimates PAF values; however, these estimates can likely be improved by better predicting the MVF capability, which is fundamental for the prediction of the PAF capability.

The ability to adequately and reliably predict PAF values depends on correctly identifying which factors affect PAF selection. Historically, research has focused on task related factors including: box size, lift height, reach distance, asymmetry, carrying method, handle type, wrist posture, obesity, etc. (Mital et al., 1989; Ciriello, 2001; Ciriello et al., 2001; Singh et al., 2009; Wu and Chang, 2010). These influential factors are then used to develop a table or regression equation, or series of equations to predict PAFs based on the associations between these factors and the PAF outcome. Lu and Aghazadeh (1994) have comprehensively reviewed this process. They concluded that more task factors need to be investigated and more participants need to be assessed to improve predictive capabilities. The current research suggests an alternate approach to improving PAF predictive ability. It advocates that attention should shift from investigating how various task factors affect PAFs towards understanding how the underlying biomechanics directly impact PAF selection (Chapter 6).

A limiting factor dependent proportionality based model is consistent with numerous published associations based on numerical regression approaches. Historically, predictive capability has been dependent on the strengths and limitations of regression based approaches (Ayoub et al., 1980; Genaidy et al., 1988). For example, Jiang and Ayoub (1987) used the predictor variable strength (calculated as a composite score) and were able to predict psychophysical lifting capacity with a variance explanation (r^2) of 0.924. Similarly, Genaidy et al. (1990) used predictor variables including: lift height, box dimension, and handle type, but not strength, and predicted psychophysical lifting capacity with variance explanations (r^2 values) of 0.83 and 0.85 for males and females, respectively. Interestingly, strength alone was a better predictor for lifting and lowering tasks. From a biomechanical perspective, MVF capacity is governed by factors including balance, strength and friction (Kerk et al., 1998; Holbein and Chaffin, 1997; Kroemer, 1974, Seo et al., 2010; Chapter 3). During lifting and lowering, when the load is close to the body, strength is more likely to limit capacity than balance or friction (Kerk et al., 1998). This may explain why strength alone had greater predictive capacity in terms of the variance in the data set, as it plays a prominent role in determining the threshold limit for capacity. However, the other predictor variables still accounted for a large amount of variability as those factors influence the demand on the body in terms of joint moments (Davis et al., 1998; Kingma et al., 2004; Kingma et al., 2006). As a result, strength may be a better predictor as it relates to the peak capacity, but external task factors are also important as they are needed to determine the demand relative to that capacity. Together both forms of information are needed to determine the relative demand on the body.

The weakest link proportionality approach to predict PAF values during one-handed exertions at once per minute is dependent on knowing the MVF capacity. Although the model

provides an underestimate of PAFs relative to experimental values, it is predominantly due to the underestimation of the MVF prior to introducing the weakest link proportionality (as shown in Chapter 4). Additionally, Kerk et al. (1994; 1998) presented a two-dimensional MVF prediction model and similarly reported underestimations up to 36% for pushes, pulls, lifts and downward presses. Since these models use similar data sets to derive the strength constraint thresholds (Schanne, 1972; Stobbe, 1982) it is expected that they would result in similar conservative estimates. The underestimation may be a result of the underlying data used to describe the constraint thresholds. The original data may not be well suited for use in predicting strengths for a modern day workforce. Research has shown secular changes in a number of health and performance related measures (Tomkinson and Olds, 2007; Danubio and Sanna, 2008). Specifically in terms of strength, Westerstahl et al. (2003) reported a significant increase in isometric strength in young males from 1974 to 1995. Further, Schanne (1972) described that the strength thresholds developed using a regression model underestimate empirical strength thresholds by approximately 30%. This is a result of the assumption that joint strengths are independent. In the underlying regression equations (Schanne, 1972; Stobbe, 1982) moment loading at adjacent articulations does not affect the strength capability of a particular muscle group. The assumption of independence may be greatly impacting the ability to adequately determine joint strength thresholds. Therefore, in order to improve MVF estimates and subsequently PAF estimates, more current research is needed regarding constraint thresholds that govern MVF capacity.

The weakest link based proportionality is attractive as a possible guiding principle to facilitate PAF predictions from both evidentiary and practical application perspectives. This research supports that proportional relationships do exist between MVFs and PAFs (Potvin et

al., 2006; Potvin, 2007; Andrews et al., 2008). However, this work additionally demonstrates that the usefulness of this proportional relationship can be improved by accounting for the biomechanical weakest link. Although the existence of this weakest link based relationship has been previously hypothesized (Nussbaum and Lang, 2005), this research in concert with the results from Chapter 6 provide quantitatively evidence for this relationship. Future work is needed to demonstrate if the weakest link principle can be more readily applied as a general rule for predicting PAFs from MVFs. Since the exertions examined here were only marginally different from those previously tested (Chapter 6); it is difficult to make a broader conclusion that the weakest link principle will extrapolate well to substantially different exertion scenarios. If the principle does extrapolate well, then a weakest link modeling approach could provide a means to forecast capacities for a number of workers in a number of jobs much more efficiently.

The outputs from the current model for predicting PAFs for unilateral exertions occurring once per minute should be interpreted with caution. As noted, the weakest link proportionality principle has not been evaluated over a wider range of tasks to support its general use. In addition, limitations and assumptions in the fundamental biomechanical model should guide interpretation. The biomechanical model is subject to errors arising from the postural data collection, such as inter-trial skin motion and marker placement accuracy, and the use of population-based anthropometric tables to estimate segment masses. These considerations can cause some uncertainty in the center of pressure calculations guiding the ability to determine if balance is limited, and the linked segment modeling outcomes guiding the ability to determine if strength is limited (Chapter 4). These effects were lessened through the use of rigid clusters to help reduce potential artifacts in the motion capture data (Kingma et

al., 1996) and the error from the use of anthropometric tables should be randomly distributed across participants. Further, only males were tested during this study. If this approach is to be applied to predict thresholds in industry, more work is needed to evaluate if the same principle applies to females. However, the use of male participants with industry experience provides support that this principle is justified for male workers performing unilateral manual materials handling tasks.

7.5 – Conclusion

This study presented and evaluated if PAFs could be predicted for unilateral pulling and downward pressing tasks by proportionally scaling predicting MVF capacity based on the corresponding biomechanical weakest link. The results indicate that this approach conservatively predicts PAFs. The underlying weakest link proportionality principle for predicting PAFs from MVFs is supported; however, overall predictive capacity could be enhanced by improving estimates of MVF capacity prior to predicting the PAF.

Chapter 8 – General Conclusions

8.1 – Revisiting the thesis statement

This dissertation aimed to address the global thesis that psychophysical strength is quantitatively related to simulated job static strength during one handed exertion tasks. Further, it aimed to determine if the relationship was related to the exposure at the weakest biomechanical link, most likely to limit maximum capacity during a task. This collection of manuscripts provided quantitative evidence to support the thesis that simulated job static strength capacity is limited by specific biomechanical limiting factors (whole-body balance, shoe-floor friction, and joint strength). Further, psychophysical strength was selected as a proportion of simulated job static strength, where the proportion was specific to the underlying biomechanical limiting factor.

8.2 – Revisiting the thesis aims and objectives

1. To advance our capacity to predict simulated job static strength by developing a biomechanically driven model to predict maximum hand forces during unilateral exertions.
 - a. *A 3D hand force prediction model (3DHFP) was developed to estimate simulated job static strength during unilateral exertions. The model included a novel approach for predicting variability in the estimates by including a probabilistic modeling component. This inclusion provides population scalability in the estimates of simulated job static strength. A model evaluation determined that it is necessary to accommodate for additional off-axis forces that re-direct the force vector away from the prescribed direction. Further, it was determined that the model conservatively predicted unilateral simulated job static strength by*

approximately 25%. The underestimation may be related to overly restrictive biomechanical constraint thresholds in determining MVF levels.

2. To predict the element, or “weakest link”, most likely to limit simulated job static strength during a unilateral exertion.

a. The 3DHFPM was successful ten times out of twelve at predicting biomechanical factors that limited unilateral simulated job static strength compared to biomechanical limiting factors determined experimentally.

3. To investigate how psychophysical strength may be determined within the strength paradigm.

a. Chapter six demonstrated a quantitative relationship between simulated job static strength and psychophysically acceptable strength. This relationship was supported in Chapter seven. This work provides quantitative evidence to support previous research (Nussbaum and Lang, 2005; Potvin et al., 2006; Potvin, 2007; Andrews et al., 2008) suggesting that psychophysical strength could be selected as a proportion of simulated job static strength.

4. To examine the plausibility of a weakest link explanation for the relationship between simulated job static strength and psychophysical strength.

a. This thesis provided insight into the proportionality relationship described as part of objective three. This body of work demonstrates that the biomechanical factor limiting job static strength, or the ‘weakest link’, also plays a role in the selection of a psychophysically acceptable force. When the weakest link is joint strength, psychophysically acceptable forces were selected at approximately 2/3rds of the simulated job static strength; however, when whole-body balance was the weakest

link psychophysically acceptable forces were selected at approximately 4/5th of the simulated job static strength.

5. To determine the feasibility of developing an ergonomic assessment tool combining the benefits of both simulated job static strength and psychophysical strength approaches (Figure 8.1 - adapted from Figure 1.2 in Chapter 1).

- a. *This dissertation outlines the development of a model to predict simulated static strength (3DHFP), the quantification of a principle that may help relate psychophysical strength to simulated job static strength, and presents an updated model to include predictions of psychophysical forces scaled from estimates of simulated job static strength based on the principle. Based on these contributions it is feasible to develop an ergonomic assessment tool combining the benefits of both simulated static strength and psychophysical strength with the following two caveats. First, the simulated static strengths predicted by the 3DHFP are well below actual capabilities, where underestimation has also been reported for other strength prediction models. This underestimate also affects the subsequent ability to predict psychophysical strength. As noted in Figure 8.1, this needs to be addressed before attempting to mobilize this approach into a useful ergonomics tool.*
- b. *Second, this thesis only examined one-handed exertions at approximately shoulder height. Additional work is required to determine: a) if the model is capable of predicting simulated static strengths at other work locations, b) if the same “weakest link” approach can be used to predict simulated strength in bilateral tasks, and c) if the weakest link proportionality principle holds across other tasks.*

c. So, although this approach seems feasible based on the research performed throughout this dissertation, at present it may not yet be practical to implement as an ergonomic assessment tool.

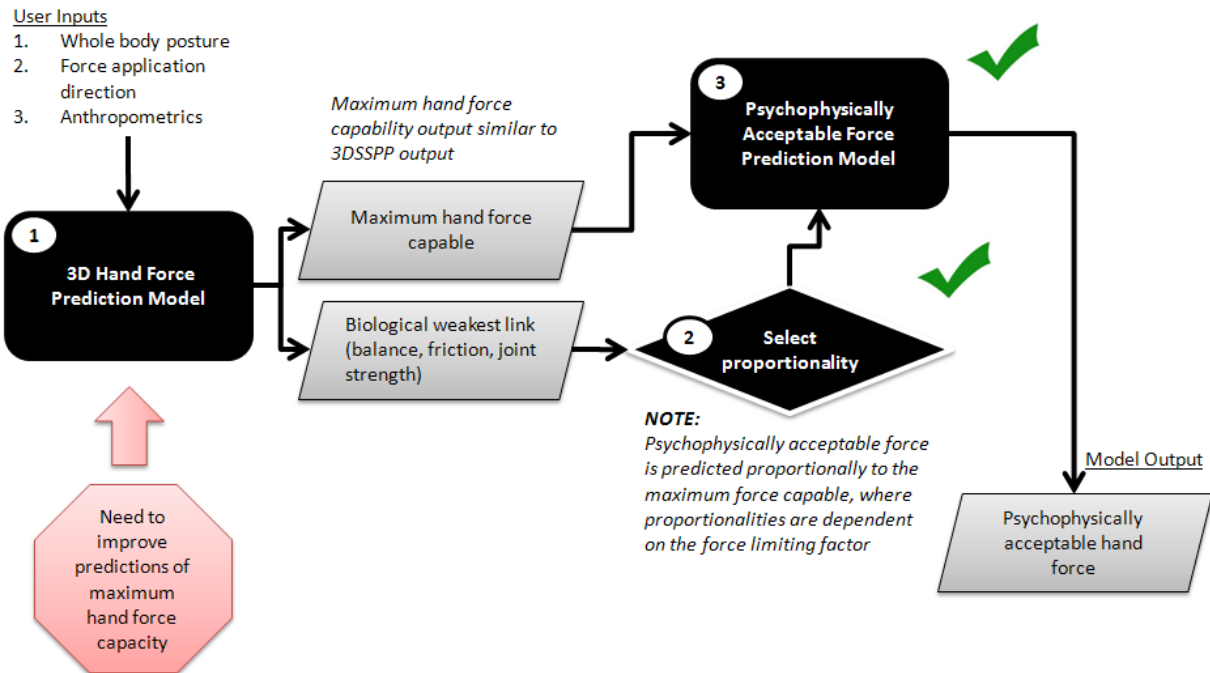


Figure 8.1 – A conceptual overview of the components and models developed throughout this dissertation (adapted from Figure 1.2). As identified in the figure, the 3DHFBPM should be improved if this approach is to be transferred into an applied ergonomics assessment tool. Within the context of this research, the proportionality and psychophysical force prediction are feasible with an improved capacity to predict maximum hand force capability.

8.3 – Novel contributions

This thesis offers two novel contributions to the field of occupational biomechanics: firstly, the introduction and use of probabilistic approach to modeling hand force capacity, and secondly, experimental findings uncovering a weakest link proportionality principle relating psychophysical strength to simulated job static strength. The probabilistic approach used to select constraint thresholds (described in Chapter 4) presents a new way to understand variability in hand force capacity. It is well known that hand force capacity is variable across

a population (Warwick et al., 1980; Haslegrave et al., 1997; Kerk et al., 1998; Das and Wang, 2004), a finding supported by the variability in force capacity observed in Chapter 3. Within the biomechanical weakest link paradigm used here the variability in hand force capacity is realized because of the underlying variability modeled in constraint thresholds. Although this thesis does not focus on this issue directly, the novel application of the probabilistic approach in this manner invites future research aimed at investigating the role of variability in underlying constraints on the variability observed in hand force capacity.

The experimental evidence demonstrating the weakest link proportionality principle advances the current idea that a uniform proportionality may exist. The novelty of this research is not related specifically to the development of the principle, but in providing a new way to look at both psychophysical and simulated static strength, revealing the principle. By further classifying these exertions based on the biomechanical weakest link, both the experimental data collected within this dissertation, and previously reported data demonstrate the same relationships (Chapter 6). This insight may provoke future research investigating why different proportionalities exist for different constraints, or why the proportion magnitudes seem so consistent within each constraint classification, across participants, although the psychophysical forces are self-selected. Addressing these issues in the future, guided by the proportional relationship demonstrated here, may help us understand the pathways through which a person arrives at a psychophysically acceptable force.

8.4 – Research significance and impact

Psychophysically derived thresholds are used extensively in applied settings, but continue to be criticized for their dependence on subjective measures (Ayoub and Dempsey, 1999; Snook, 1999; Dempsey, 2006). This dissertation has identified the potential for a

quantifiable biomechanical explanation linking psychophysical forces and maximum capable forces. This relationship is based on Newton's third law, where every action has an equal and opposite reaction. Under that law, in the case of maximum exertions, the body can only apply increasing levels of force until the corresponding reaction effect is greater than the body can resist, in terms of balance, friction, or strength. Identifying that psychophysical forces are in part dictated by this same explanation is a significant contribution. This dissertation demonstrates that participants apply psychophysically acceptable forces such that the reaction response reaches a pre-determined margin, below the absolute threshold limit to maintain balance or fall within joint strength capability. This margin is related specifically to the threshold most likely to be exceeded, be it balance or joint strength.

8.5 – Limitations

A weakest link proportionality principle is posed throughout this dissertation. There are a number of notable cautions that should inform its future use and interpretation.

8.5.1 – Determining the weakest link

First, the approach for determining a “weakest link” must be refined. Based on the results of the model evaluation (Chapter 5) it was clear that predictions were significantly lower than observed values. This may be a result of the quality of data being input into the model, the weakest link model logic, or the constraint thresholds selected from previous literature.

In terms of data quality, caution was used to ensure appropriate procedures and protocols were followed to determine representative posture data. The most critical limitation of the procedures used was discussed in Chapter 6, stating the limitations in using a single cluster placed over the sternum to determine virtual markers representing landmarks all over

the trunk. Although this approach was not ideal, the sternum was selected as the most rigid point on the trunk that would not be covered up when the bracing apparatus was used. In terms of the model logic, a weakest link approach was used, which is predicated on Newton's third law. The caveat here is that the equal and opposite reaction must not cause balance to be lost or joint strength to be exceeded. The evaluation data presented in Chapter 5 indirectly speaks to this issue. If a weakest link approach applies, then individuals would be more likely to adjust the force direction from a prescribed direction (e.g. pulling horizontally and down during a horizontal pull) to protect against a weakest link. In comparing the differences between actual and prescribed force application directions, this rationale seems to hold true, at least for downward exertions. Using a weakest link approach, participants did manipulate the force direction to protect against joint strength limiters, supporting the weakest link logic used in the model.

The use of previously published data remains a target for understanding the conservative predictions. With reference to joint strength limits, the work of Stobbe (1982) and Schanne (1972) still likely represents the most comprehensive description of individual joint strengths collected on the same populations using the same methodology. However, there are several limitations that they describe in their seminal work. Among those limitations, Schanne (1972) describes that predicted hand forces are typically underestimated by approximately 30% during one-handed exertions, when using the regression equations to determine joint strength capacity. Extending from this Schanne (1972) describes that this undershoot is dominated by a faulty assumption that moment loading at adjacent articulations does not affect the strength capability of a particular muscle group. Schanne (1972) effectively points out that joint strengths are treated as independent, though they are quite dependent. This work has also

relied on this independence assumption, which is a limitation that may be responsible for the model underestimations during strength limited exertions.

Further, strength attributes may differ between participant pools. The original data from Schanne (1972) was established using university students, while the data generated later by Stobbe (1981) was based on a different sample, possibly with different force producing attributes. Figure 8.2 illustrates the maximum strengths observed between the university students (Study 1) with the workers hired (Study 3). Qualitatively, the strengths are similar for most exertions; however they do differ significantly in the free “FFP” down exertion and during the medial and pulling exertions while braced “UBB”. The weakest link modeling approach assumes that these maximum forces are all governed using the same population joint strength limits. This assumption may therefore be impacting the underestimations; however, given the similarity in maximum strength between the groups during six of the nine different exertions, it is unlikely that this assumption is a dominant explanation for the underestimations made by the model.

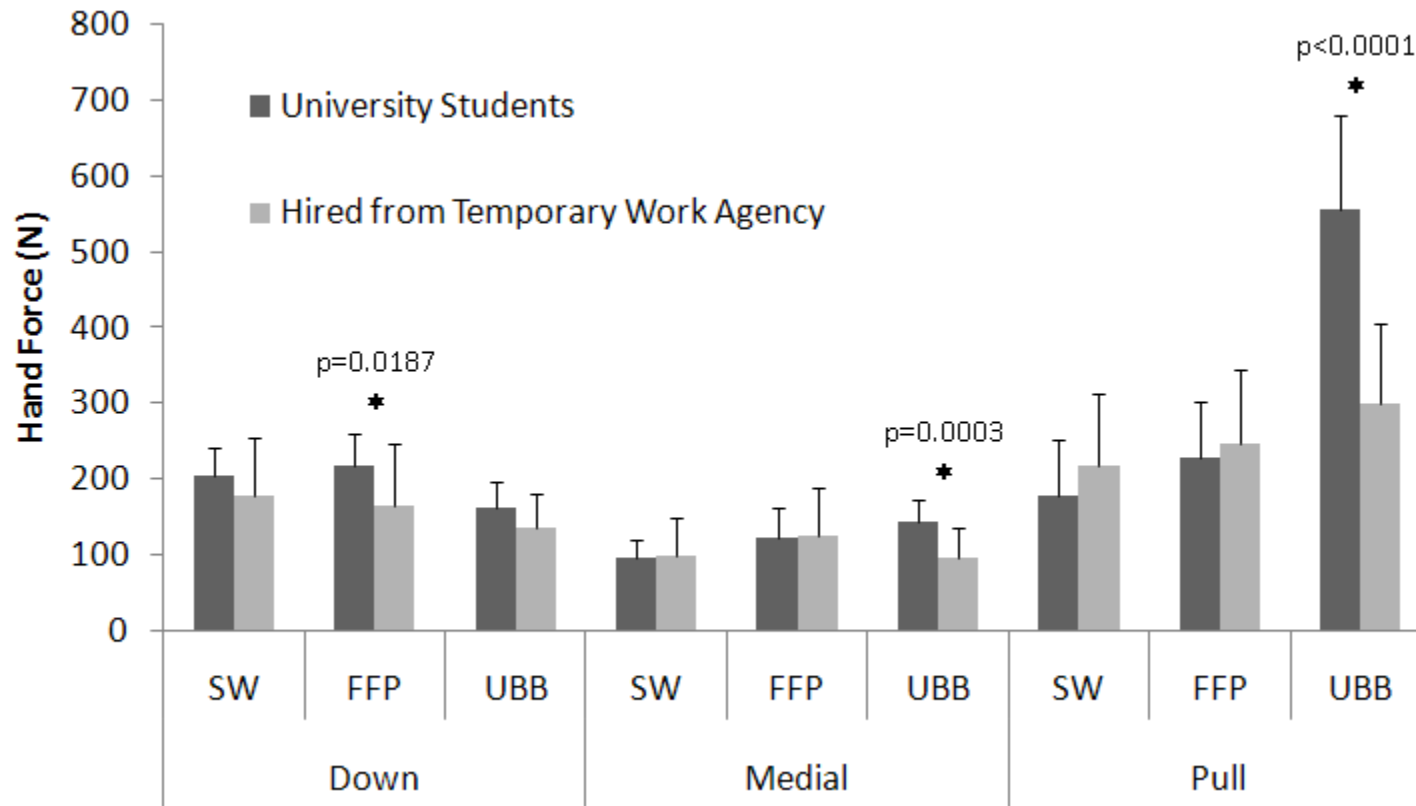


Figure 8.2 – A comparison of maximum hand forces produced by male university students (Study 1) and males hired from a temporary work agency (Study 3). The asterisks ‘*’ denotes significant differences between the strengths of university students and workers hired from a temporary work agency within the respective conditions.

Additionally, the data used to determine balance thresholds may also have limitations that may further limit the predictive capability of the model described in this work. Holbein and Chaffin (1997) discuss how the use of static limits may underestimate the actual center of pressure (COP) excursion limits, especially when one considers the ranges of excursions that are present under more dynamic conditions. However, the relatively static postures tested throughout this work likely mitigate the effect of this limitation. More importantly, Holbein and Chaffin (1997) also noted that the COP excursion limits they reported using an un-loaded and loaded leaning protocol, were less than those observed by Kerk et al., (1998) using a two-handed exertion protocol. For example, participants may move their COP less when leaning versus when pushing or pulling. Although the data presented by Kerk et al. (1998) demonstrated greater COP excursion thresholds, their data set provided only anterior and posterior limits in a shoulder width foot position. The data presented by Holbein and Chaffin (1997) were more restrictive, but included anterior, posterior and lateral limits for both shoulder width and asymmetric foot placements. It remains as a limitation in this work that the more conservative limits were chosen to represent the COP excursion limits (from Holbein and Chaffin, 1997), when more task specific limits were available (Kerk et al., 1998). This choice was predicated on the need to have more than only anterior and posterior limits in the three-dimensional case.

8.5.2 – The specificity of the task

The second notable caution is the specificity of the task used to uncover the weakest link principle. The proportional relationship was demonstrated using one handed exertions completed at once per minute, for thirty minutes. Although Chapter 6 demonstrated that this

relationship might exist across other types of exertions, it remains crucial to experimentally demonstrate consistency in this principle across a wider range of exertions or tasks. Also, this relationship is dependent on psychophysical exertions completed at once per minute with low duty cycle (2% duty cycle). Psychophysically acceptable force estimates are sensitive to work time, and specifically the duty cycle (Moore and Wells, 2005). It would be expected that as the duty cycle increases, the psychophysically acceptable forces would decrease. Therefore, it is important to understand that this principle may be task specific, but more importantly duty cycle specific.

8.5.3 – Postural consistency

The third notable caution relates to postural consistency. As discussed in Chapter 6, the weakest link is determined based on the posture assumed during the MVF trial. Although the gross posture remained consistent (foot locations, hand locations) the participant may have subtly altered specific joint positions within the context of the gross posture when performing a psychophysically acceptable force trial. Although specific positional differences were not quantified, when the actual psychophysical trials were classified based on the weakest link, in nearly all cases the COP limits were reached. Since the hand forces were lower than those measured during the MVF trials the joint moments were considerably lower by default, but it would be expected that the COP excursion would also be less. Using the description provided in Chapter 4, and illustrated in Figure 8.3, the following explanation is provided.

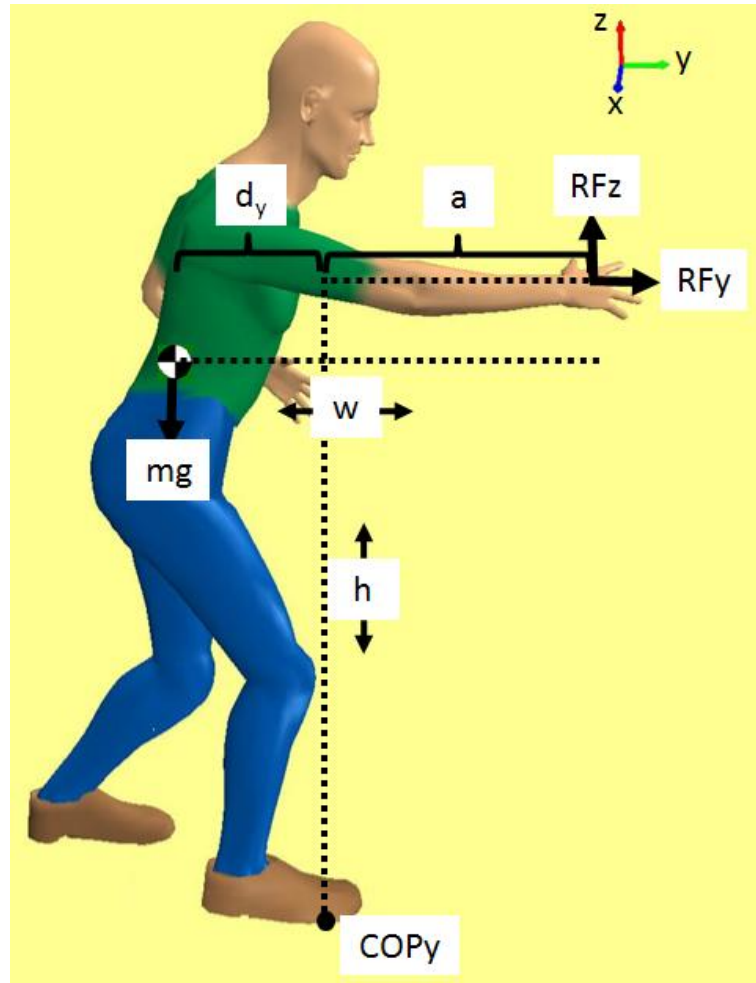


Figure 8.3 – A reproduction of Figure 4.7 - Sagittal plane diagram used to demonstrate how the static moment equilibrium about the COP in the global X axis was determined. The specific variable definitions are described with the original Figure 4.7. *Note: the hominoid graphic was generated using the 3DSSPP™.

During MVF trials, participants chose to shift their hips and consequently their center of mass (COM) in a posterior direction, increasing the moment arm between the COM and COP to accommodate a higher moment produced by the hand force. During psychophysical trials, the hand force was lower, and the corresponding rotation moment it caused was lower. Concurrently, participants did not shift their hips and COM as much, reducing the moment arm and subsequent moment generating capacity. This was not quantified directly, but was

observed as the vertical projection of the COM and the COP were displayed on plots of the base of support similar to Figure 5.5.

Although gross body postures were maintained, specific joint postures were different. This is a limitation as the same exact postures were not likely used to perform both exertions. However, in the context of the weakest link, it is assumed that participants would self-adjust their posture within the context of the gross postural requirements to enable them to produce a maximum force (Hoffman, 2008). In the posture required to exert maximally they may be at an increased risk of falling or overexerting, though it is deemed reasonable to achieve a maximum effort. However, when the force is lower, at a psychophysically acceptable level, the self-adjusted posture may not need to be as extreme, and a more conservative approach is sought. This rational remains speculative at this point, and the postural inconsistencies remain as a limitation of this work.

These broader limitations provide context for interpreting the results presented here.

8.6 – Future research

Two next steps are needed to help mobilize this research into a practical applied tool: 1) improve the ability of the model to predict maximum voluntary hand forces, and 2) validate the weakest link proportionality principle over a wider range of tasks and for a female population. Pending the outcome of these first two steps, a third step can help expand the scalability of predictions.

Model predictions can likely be improved through addressing some current model limitations. First, the inclusion of a routine to predict the hand force vector from a user prescribed hand force vector would generate more realistic moments. Optimization may provide one avenue to achieve this goal; however, more research is needed to determine why

off-axis forces are produced (de Looze et al., 2000; Granata and Bennett, 2005; Hoffman et al., 2007). Secondly, the constraint thresholds should be improved. In terms of balance thresholds, previous research has not identified multi-direction COP boundaries during forceful exertions with different foot positions (Kerk et al., 1998). Additionally, research examining the interdependence of joint strengths may help eliminate the conservative nature of those thresholds.

The weakest link proportionality principle represents a significant contribution to the field of occupational biomechanics. However, this finding should be validated and supported by additional research. The principle is underpinned by the notion that every exertion is limited by a biomechanical factor. As a result, any exertion can be classified based on the biomechanical factor by solving for equilibrium, within the confines of balance, strength, and friction. A subsequent psychophysically acceptable force, at one exertion per minute, can be determined accordingly and the proportionality principle can be tested. As is indicated in the discussion in chapter 6, different joints may also have different proportionalities. Specifically, finger joint strength may be lower than the $2/3^{\text{rds}}$ presented here. Additionally, these data are representative of a small sample of males with some manual materials handling experience. Additional work is warranted to examine if this trend holds across different sample populations, and if it transfers to a female population.

The third step is related to improving the temporal scalability of model predictions. Assuming the predictive power of the model can be adequately improved, and that the proportionality principle remains consistent, an approach is needed to scale psychophysical predictions to encompass different work frequencies and duty cycles. Potvin (2007) has suggested that a logarithmic curve fit approach may be useful to scale psychophysically

acceptable force according to duty cycle and frequency. Improved temporal scalability would increase the range of tasks for which the model could be used to predict both MVF and psychophysically acceptable forces in the workplace. However, this remains a substantially longer term goal, and its pursuit is predicated on the prior successful improvement of model predictions and more universal validation of the proportionality principle.

Appendices

Appendix 1 – Psychophysical instructions (modified from Snook et al., 1995)

YOUR JOB IS TO PUSH ON THE HANDLE EVERY TIME YOU HEAR THE BEEP AND ADJUST HOW HARD YOU PUSH ACCORDING TO THE GUIDELINES BELOW:

Instructions for choosing the force:

We want you to imagine that you are on piece work getting paid for the amount of work that you do, but working a seven hour shift that allows you to go home without unusual discomfort in the hands, wrists, forearms, neck, or shoulders. In other words, *we want you to work as hard as you can without straining your hand, wrist, forearm, neck, or shoulders.*

Please feel free to adjust the amount you are pushing with as often as you would like. We want to find out *how hard you can push without straining yourself.* Adjusting your own resistance is not an easy task. Only you know how you feel. *Do not be afraid to make adjustments.* You have to make enough adjustments so that you get a good feeling for what is too hard and what is too easy. You can never make too many adjustments, but you can make too few.

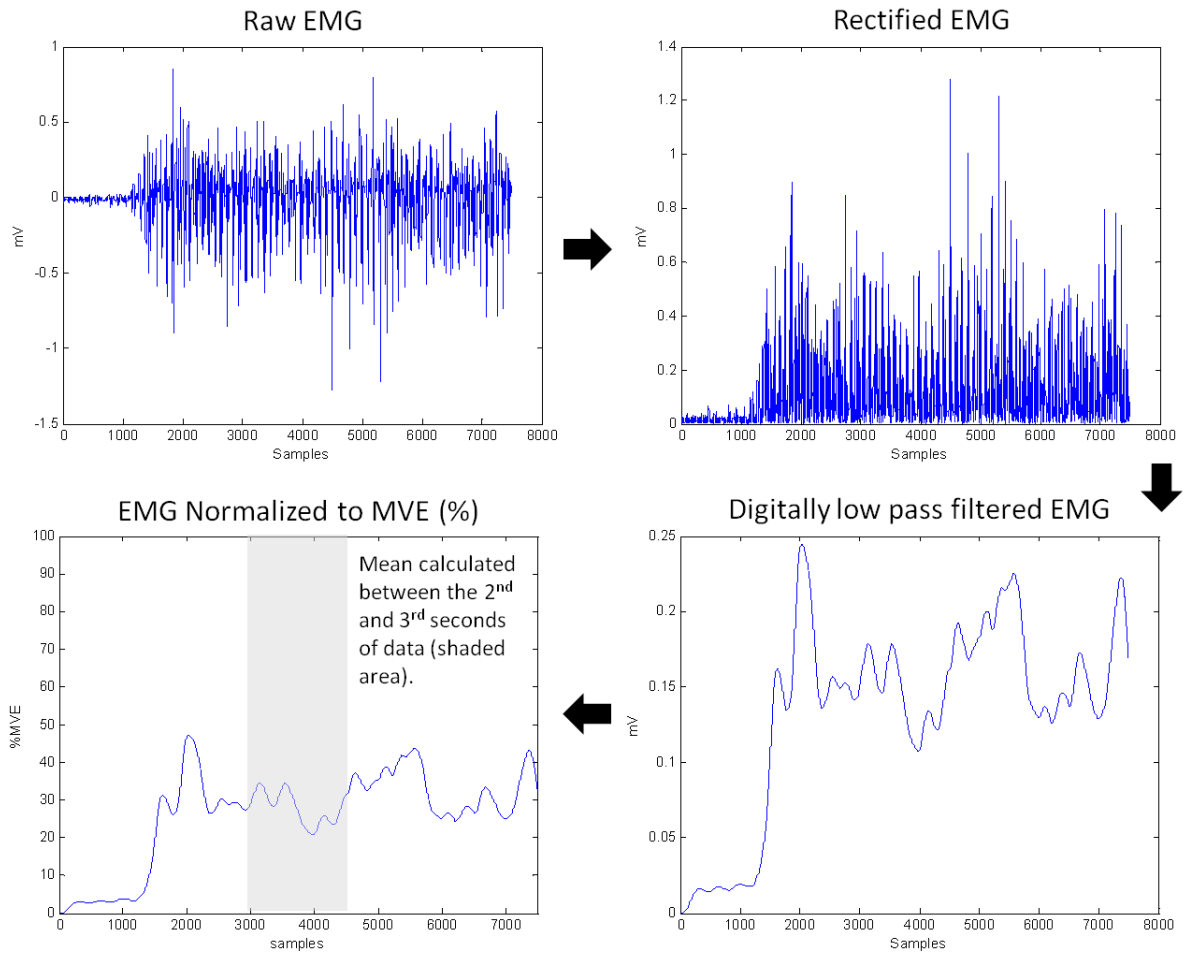
REMEMBER, THIS IS NOT A CONTEST. EVERYONE IS NOT EXPECTED TO PUSH ON THE HANDLE WITH THE SAME AMOUNT OF FORCE.

Appendix 2 – A summary of previous work experiences of the participants hired to complete the psychophysical research

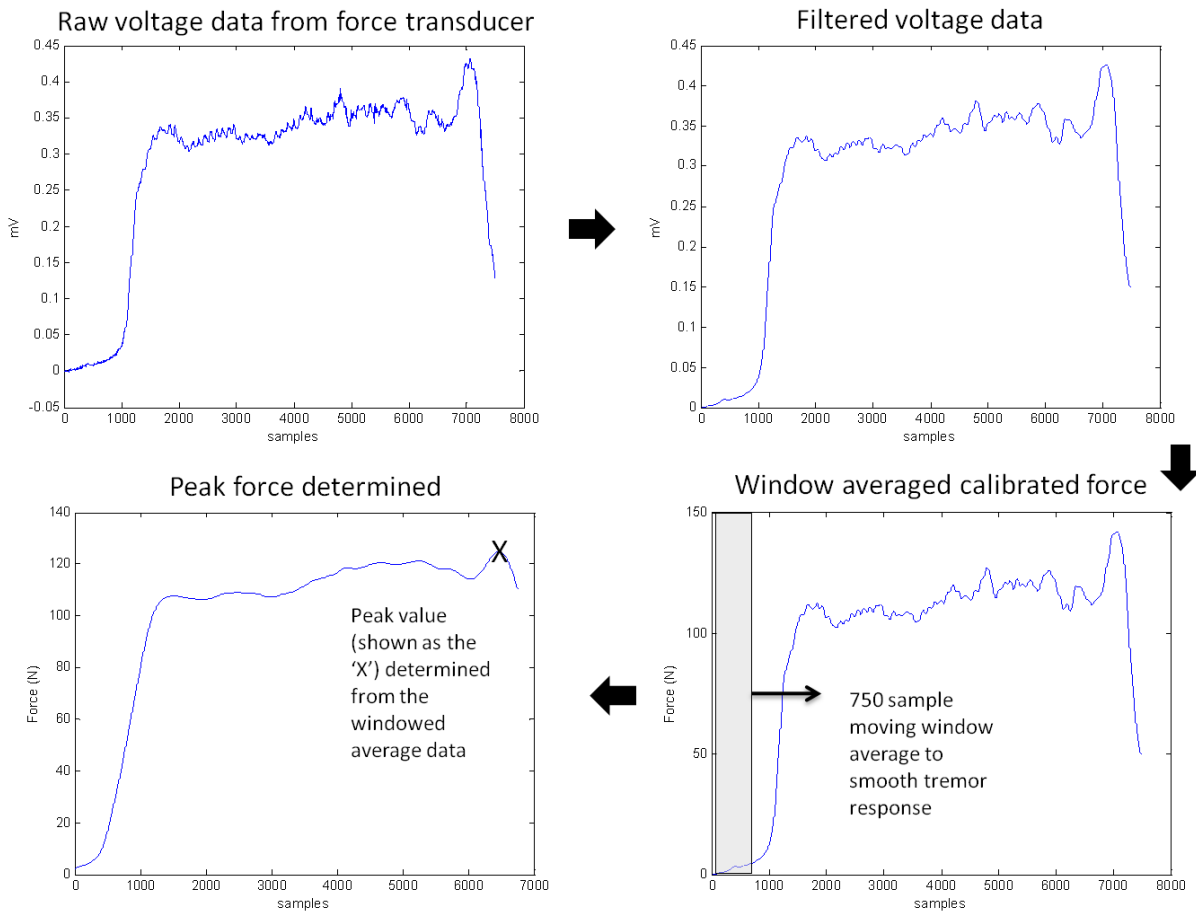
studies

Participant Code	Age	Height (cm)	Weight (kg)	Previous work experience	Year of Employment	Previous work experience	Year of Employment 2	Previous work experience	Year of Employment 3	Previous work experience	Year of Employment 4
MPA	41	183	93	Business Intermediary	May-10	Safe Harbour	Jun-10	Financial Advisor	Feb 04 - Mar 09		
ELA	58	165	79.4	Paint Line	Jul-10	Shelf Stocker	Dec-08	Leak Tester	Oct/07- Oct/08	Cleaner	Feb 06 - Oct 07
JGR	20	178	75.7	Paint Line		Dye Dry Department		Paint Department		Parts Department	
STT	48	160	63.5	General Labourer	2006 - 2009	Quality Auditor	2005 - 2006	Quality Technician	2000 - 2005		
SSM	37	183	74.8	Door Technician	Sept 08- Sept 10	Audio Technician Assistant	2005-2008				
AGR	56	160	64.4	Canadian Tire Renovator/ Organizer	Aug 08 - Nov 09	Mould Technician	2005 - 2008	Mould Technician	2005 - 2008		
CEB	60	167	80	Engine Assistant	2005 - 2010						
PKR	56	175	86.2	Bauer Industries Press Operator	Sep 1980 - Feb 2009	Press Operator	1970s				
ADA	20	185	90.7	Retail Cashier/Stocker	Dec 09 - Apr 10	Machine Operator	Jul 09 - Aug 09	Construction/ Demolition	Jun 09 - Jul 09	Retail	Mar 07 - May 08
BHU	51	171	72.6	General Labourer	2005 - 2010						
GSA	46	175	102	Industrial Spray Painter	1980 - 2010						
WCR	29	178	72.6	Kitchen Pallet Repairer	Nov 09 - May 10	General Labourer	2008	Computer Chair Assembler	2005 - 2007		
KST	44	183	104.3	Electronics Assembler	Jan/09- Apr/10						
JHE	22	175	56.7	Line Worker	Mar 06 - Mar 09	Sanitation Labourer	Mar 09 - June 10				
TLC	22	165	79.4	First Aid Instructor	Sept 06 - present	Property Manager	Sep 2008 - present	Proctor	Sep 06 - present	Assembly Line	May 07 - Aug 07
BMC	39	175	86.2	Merchandiser	Jun-10	Inspector	Nov 09 - Feb 10	Labourer	Jun-09		
PAK	51	173	81.6	Tire Maker	Oct 77 - Jul 2010	Warehouseperson	Nov 07 - Present	Groundskeeper	Apr 07 - Nov 07		

Appendix 3a – A schematic illustration of the processing steps applied to the EMG data in study 3. This data was taken from a representative participant and the EMG is from the pectoralis sternal insertion during a medial exertion in a free foot posture condition.



Appendix 3b – A schematic illustration of the processing steps applied to the voltage data in study 3 to determine hand forces. This data was taken from a representative participant and following data was acquired from a medial exertion in a free foot posture condition.



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